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(54) **SWITCHING STRUCTURES FOR HEARING AID**

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See application file for complete search history.

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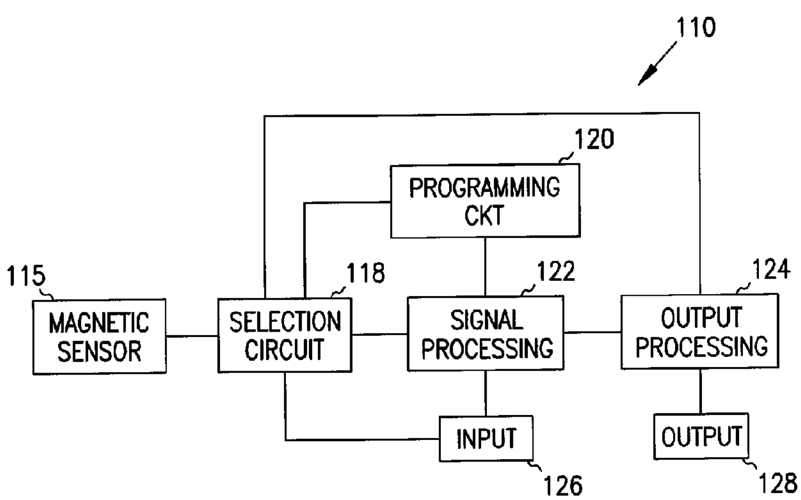
(57) **ABSTRACT**

A hearing aid is provided with a switch that automatically, non-manually switches at least one of inputs, filters, or programmable parameters in the presence of a magnetic field.

19 Claims, 12 Drawing Sheets



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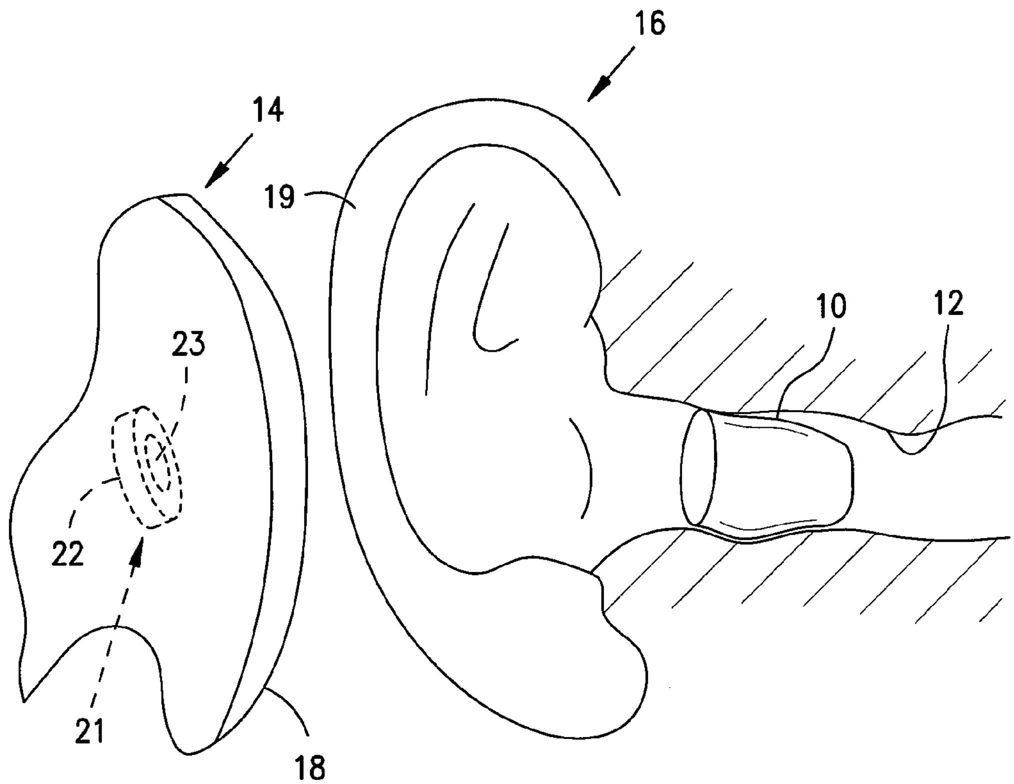
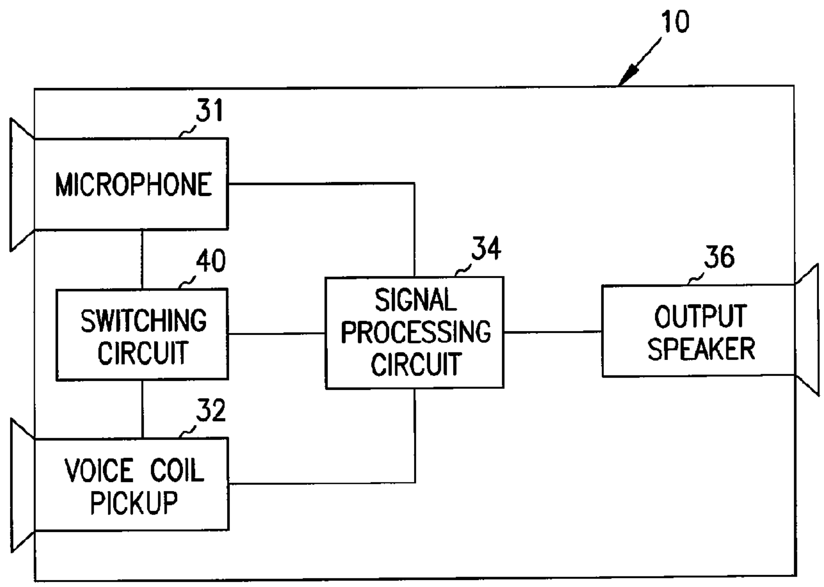
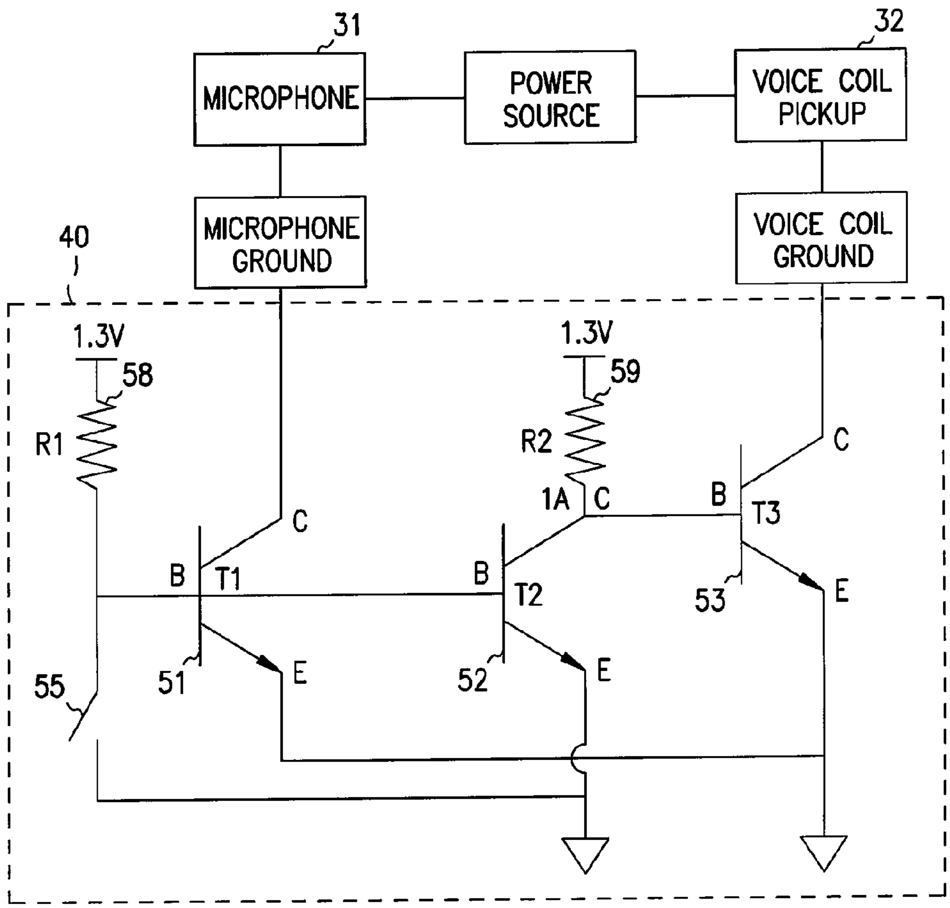


Fig. 1



10
31
MICROPHONE
40
SWITCHING CIRCUIT
34
SIGNAL PROCESSING CIRCUIT
36
OUTPUT SPEAKER
32
VOICE COIL PICKUP

Fig.2



31
MICROPHONE
POWER SOURCE
32
VOICE COIL PICKUP
MICROPHONE GROUND
VOICE COIL GROUND
40
1.3V
58
R1
1.3V
59
R2
1A
C
B
T1
C
B
T2
C
B
T3
E
E
E
51
52
53
55

Fig.3

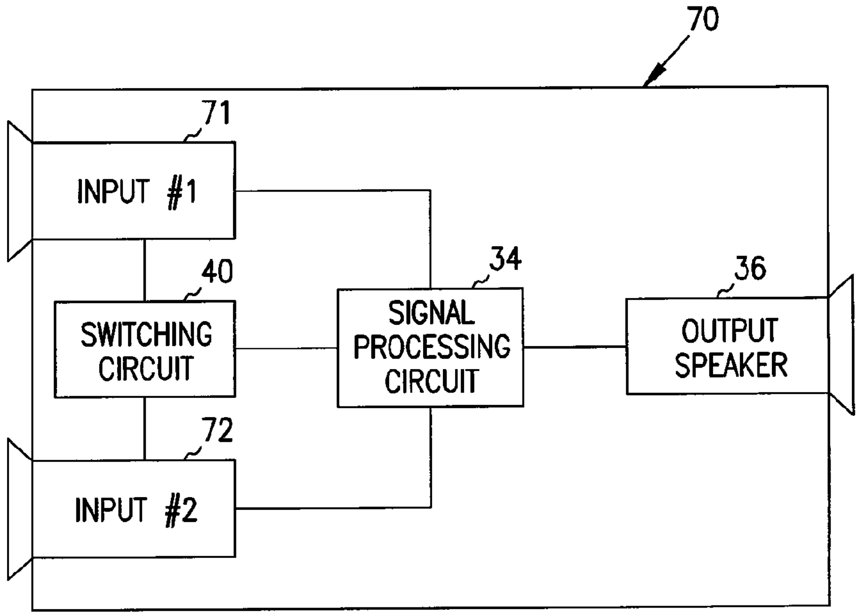

70
71
INPUT #1
40
SWITCHING CIRCUIT
34
SIGNAL PROCESSING CIRCUIT
36
OUTPUT SPEAKER
72
INPUT #2

Fig. 4

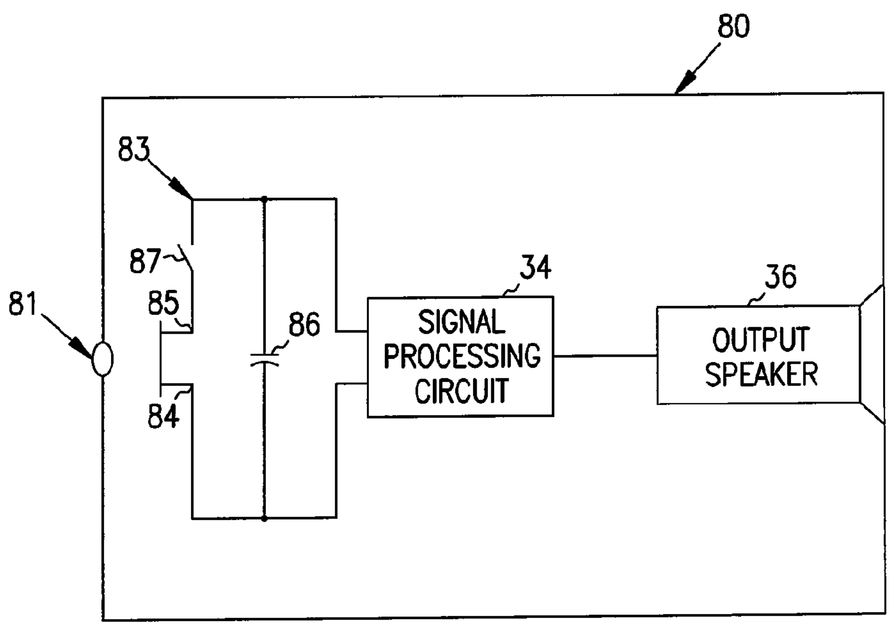

80
83
87
85
81
86
84
34
SIGNAL PROCESSING CIRCUIT
36
OUTPUT SPEAKER

Fig. 5

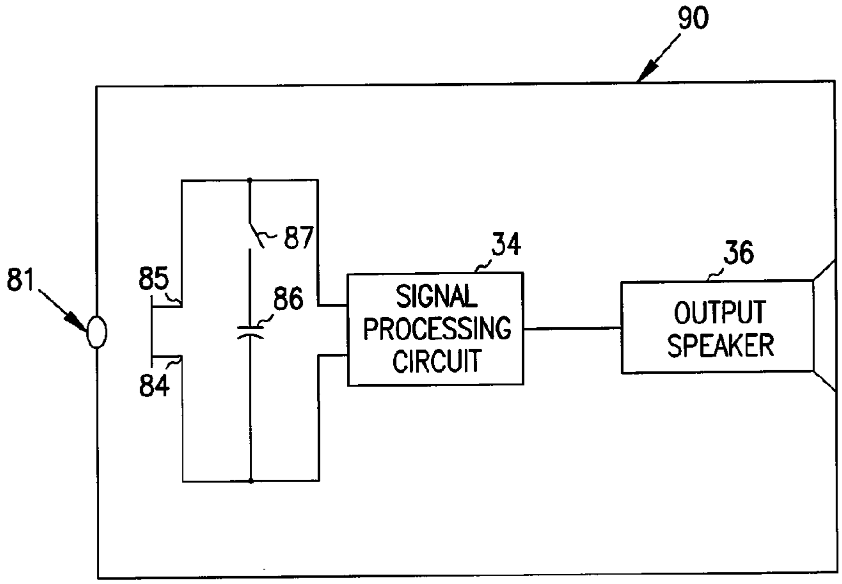

90
81
85
84
87
86
34
SIGNAL PROCESSING CIRCUIT
36
OUTPUT SPEAKER

Fig. 6

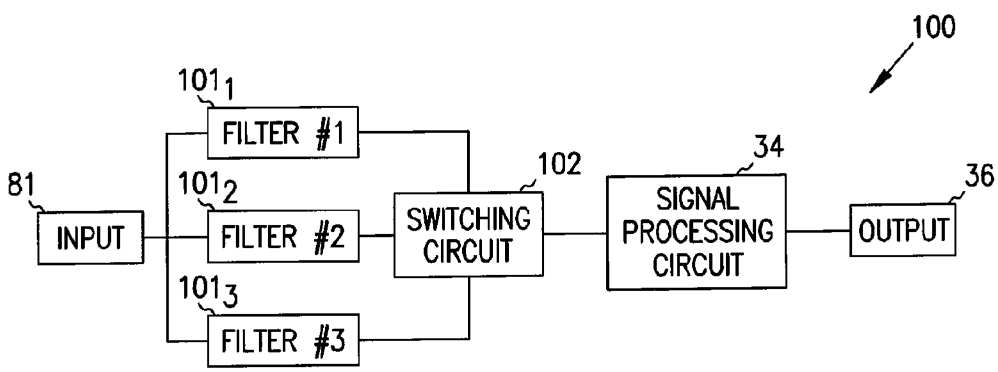

100
81
INPUT
101_1
FILTER #1
101_2
FILTER #2
101_3
FILTER #3
102
SWITCHING CIRCUIT
34
SIGNAL PROCESSING CIRCUIT
36
OUTPUT

Fig. 7

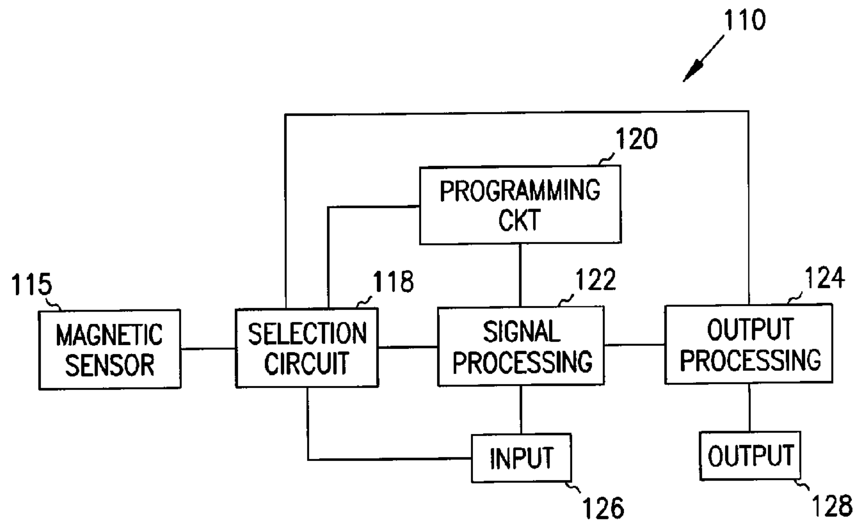

110
120
PROGRAMMING CKT
115
MAGNETIC SENSOR
118
SELECTION CIRCUIT
122
SIGNAL PROCESSING
124
OUTPUT PROCESSING
INPUT
126
OUTPUT
128

Fig. 8

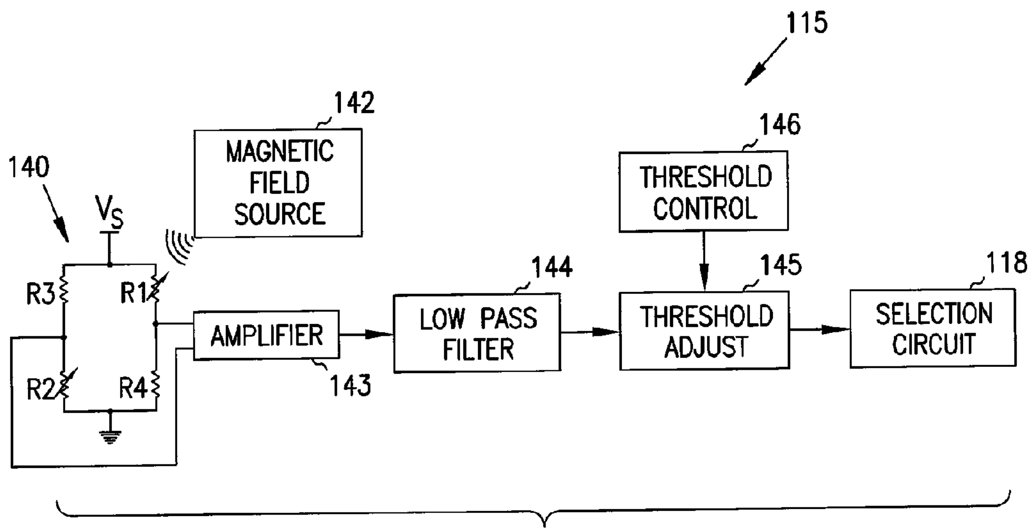

115
142
MAGNETIC FIELD SOURCE
140
VS
R3
R1
R2
R4
AMPLIFIER
143
144
LOW PASS FILTER
146
THRESHOLD CONTROL
145
THRESHOLD ADJUST
118
SELECTION CIRCUIT

Fig. 9

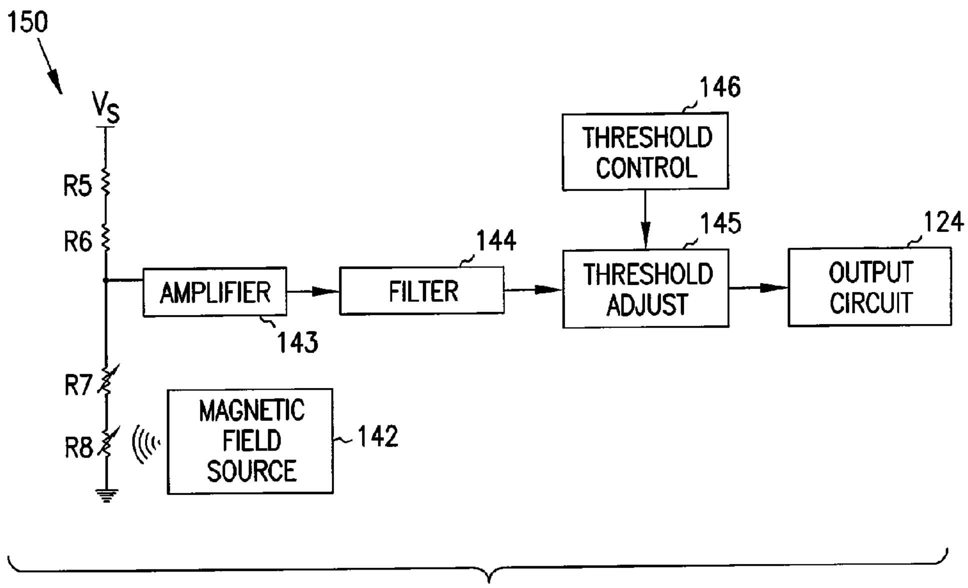

150
V_S
R5
R6
R7
R8
AMPLIFIER
143
FILTER
144
THRESHOLD CONTROL
146
THRESHOLD ADJUST
145
OUTPUT CIRCUIT
124
MAGNETIC FIELD SOURCE
142

Fig. 10

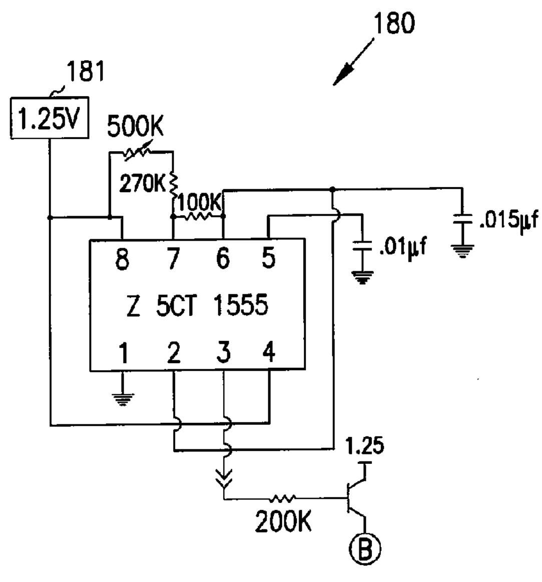
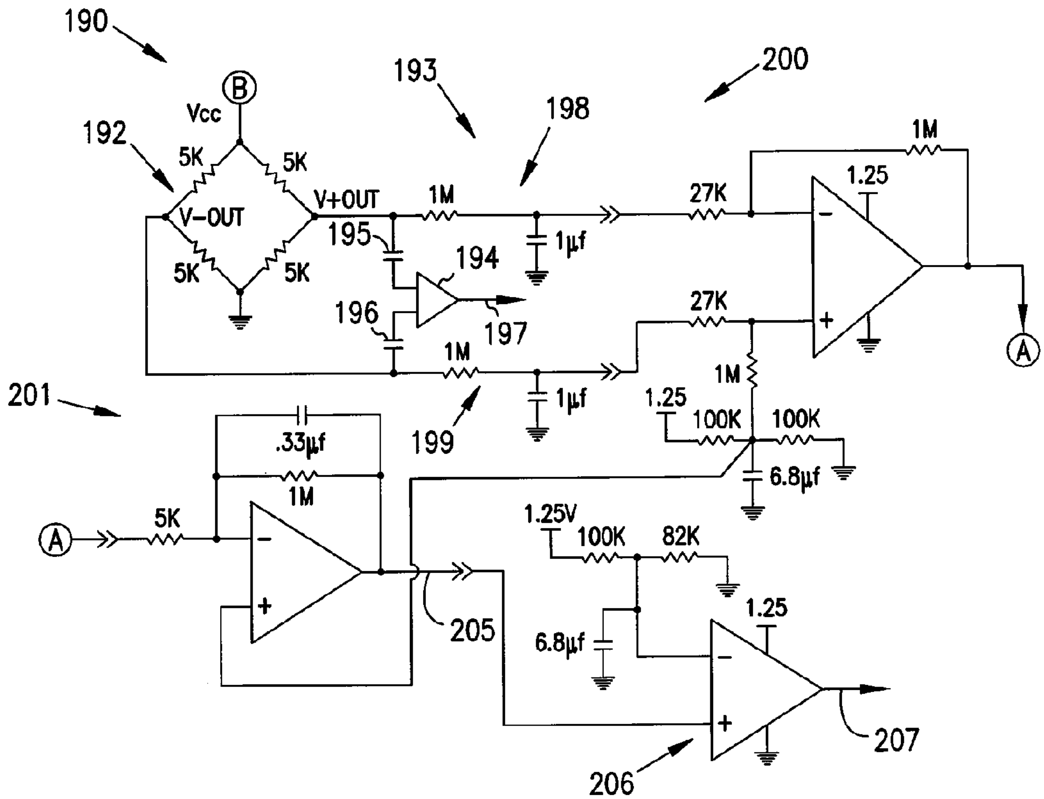
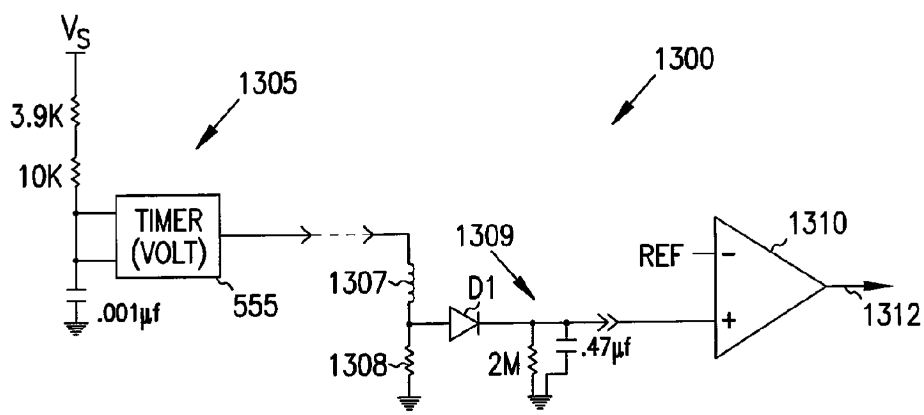

180
181
1.25V
500K
270K
100K
.015µf
.01µf
8 7 6 5
Z 5CT 1555
1 2 3 4
1.25
200K
B

Fig. 11



190
192
Vcc
B
5K
5K
5K
5K
V−OUT
V+OUT
193
198
200
1M
1M
27K
27K
1.25
195
194
196
197
1µf
1µf
1M
1M
A
201
199
1.25
100K
100K
6.8µf
.33µf
1M
5K
A
205
1.25V
100K
82K
1.25
6.8µf
207
206

Fig. 12



V_S
3.9K
10K
1305
1300
TIMER (VOLT)
555
.001µf
1307
1308
1309
D1
2M
.47µf
REF
1310
1312

Fig. 13

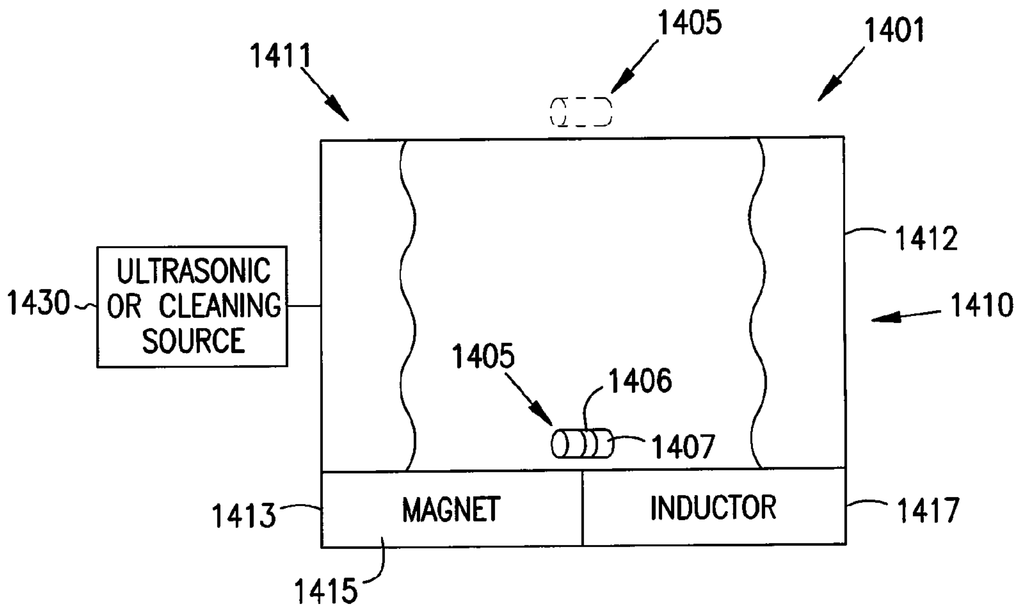

1411
1405
1401
1412
1410
1430
ULTRASONIC OR CLEANING SOURCE
1405
1406
1407
1413
MAGNET
INDUCTOR
1417
1415

Fig. 14

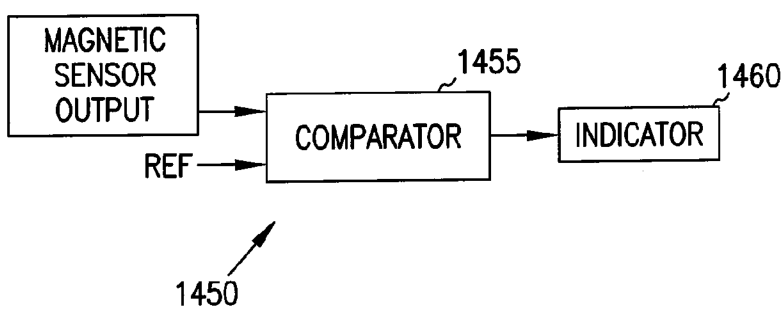

MAGNETIC SENSOR OUTPUT
REF
1455
COMPARATOR
1460
INDICATOR
1450

Fig. 15

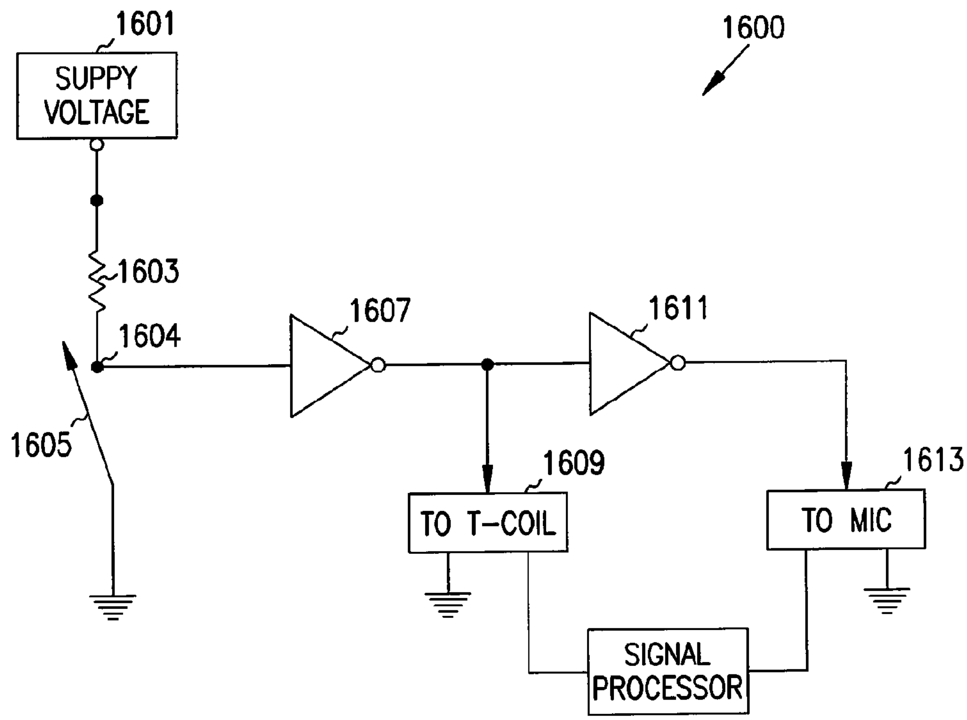

1601
SUPPY VOLTAGE
1600
1603
1604
1607
1611
1605
1609
1613
TO T-COIL
TO MIC
SIGNAL PROCESSOR

Fig. 16

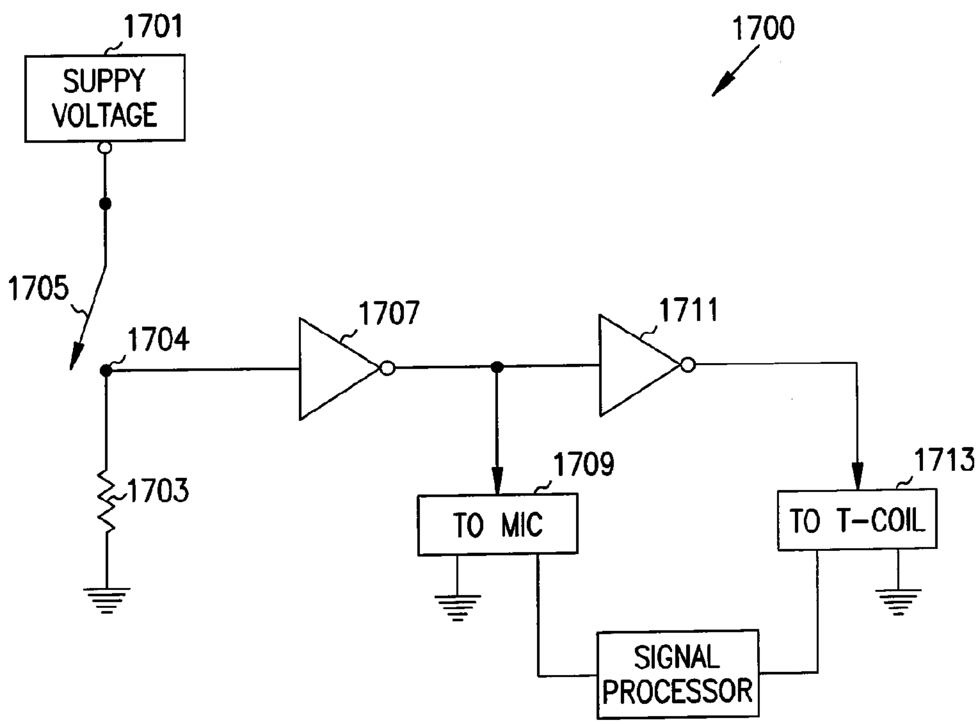

1701
SUPPY VOLTAGE
1700
1705
1704
1707
1711
1709
1713
1703
TO MIC
TO T-COIL
SIGNAL PROCESSOR

Fig. 17

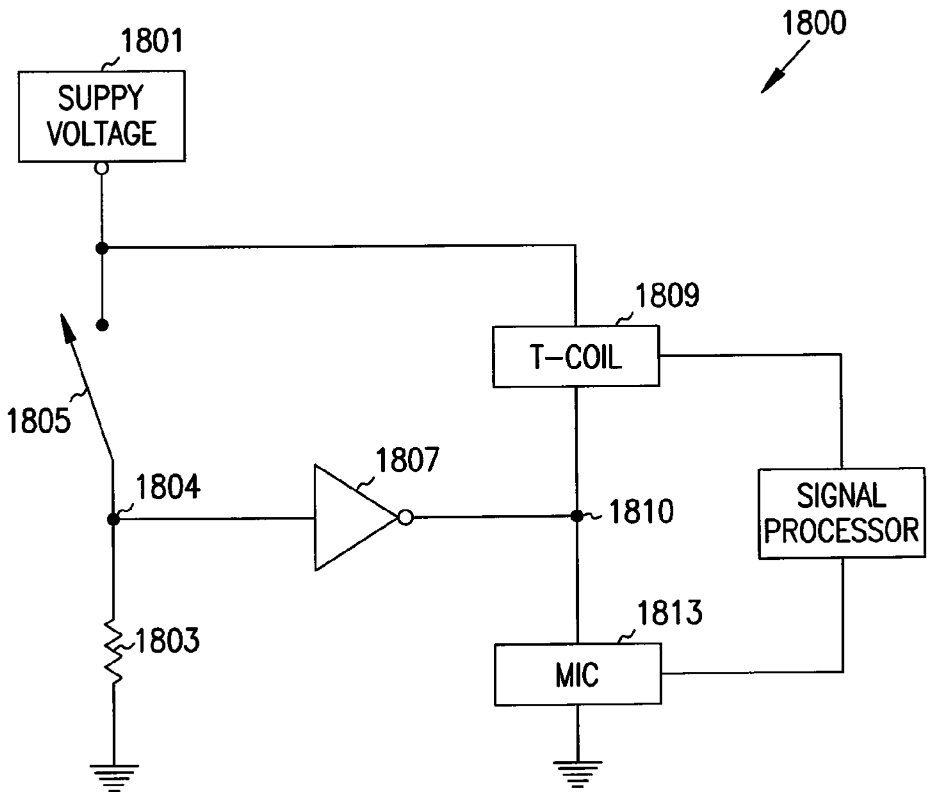

1800
1801
SUPPY VOLTAGE
1809
T-COIL
1805
1807
1804
1810
SIGNAL PROCESSOR
1813
1803
MIC

Fig. 18

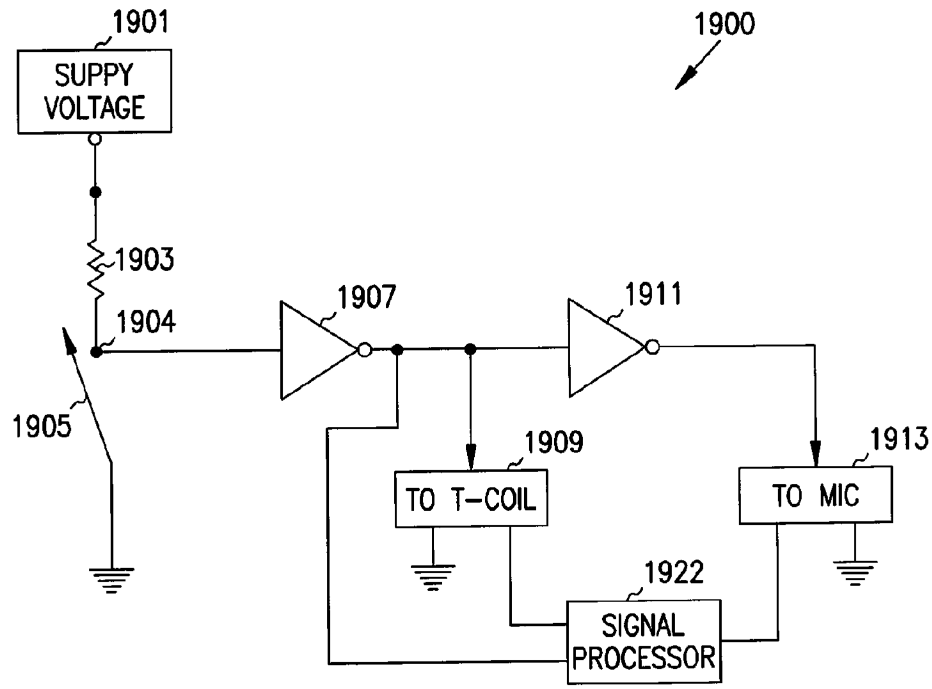

1900
1901
SUPPY VOLTAGE
1903
1904
1905
1907
1911
1909
TO T-COIL
1913
TO MIC
1922
SIGNAL PROCESSOR

Fig. 19

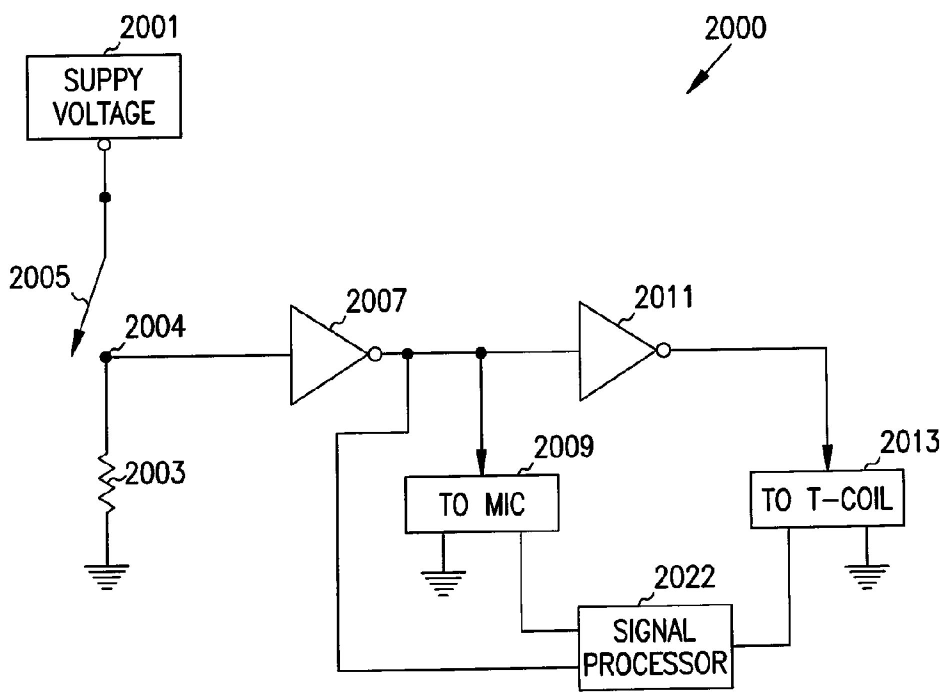

2000
2001
SUPPY VOLTAGE
2005
2004
2003
2007
2011
2009
TO MIC
2013
TO T-COIL
2022
SIGNAL PROCESSOR

Fig. 20

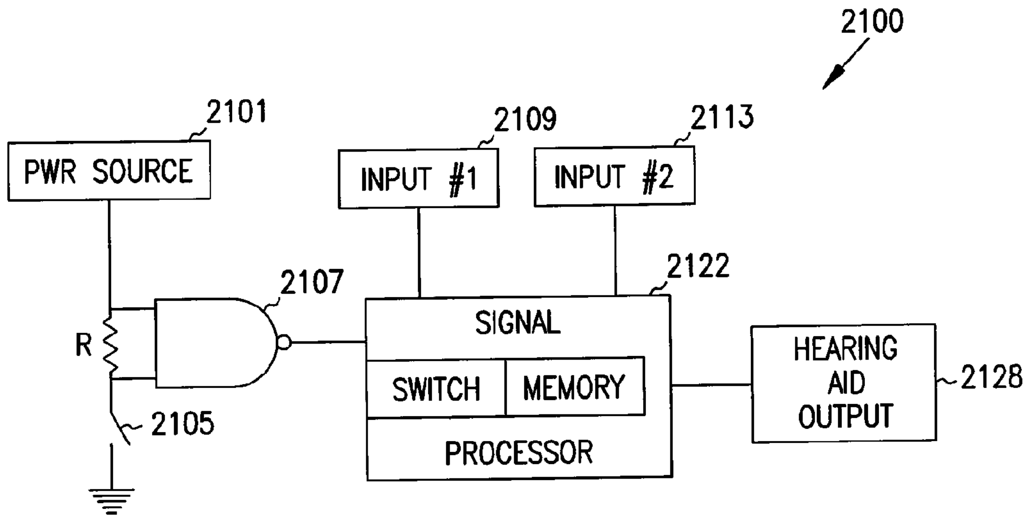

2100
2101
PWR SOURCE
2109
INPUT #1
2113
INPUT #2
2122
2107
R
SIGNAL
SWITCH
MEMORY
PROCESSOR
HEARING
AID
OUTPUT
2128
2105

Fig.21

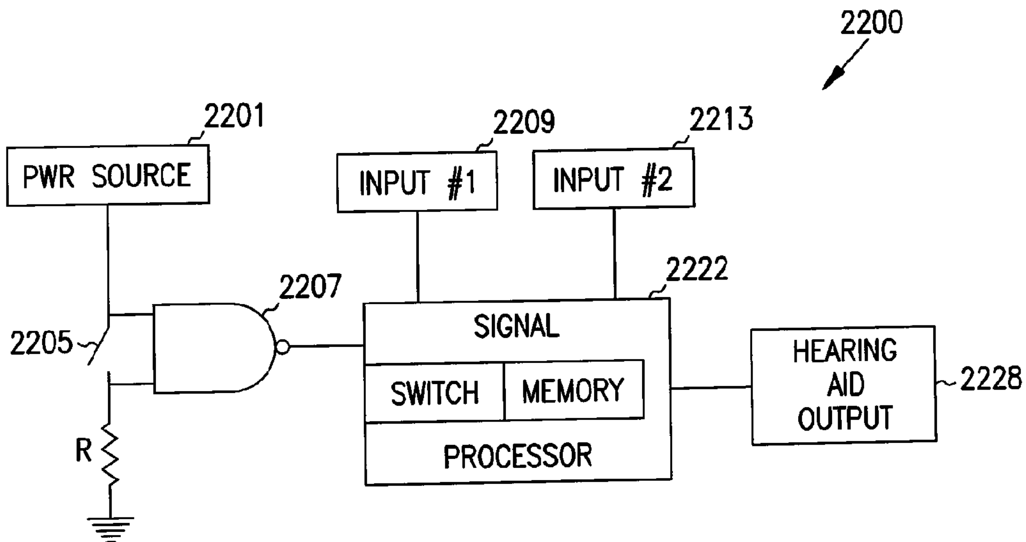

2200
2201
PWR SOURCE
2209
INPUT #1
2213
INPUT #2
2222
2207
2205
SIGNAL
SWITCH
MEMORY
PROCESSOR
HEARING
AID
OUTPUT
2228
R

Fig.22

SWITCHING STRUCTURES FOR HEARING AID

RELATED APPLICATIONS

This application is a continuation of U.S. application Ser. No. 12/107,643, filed Apr. 22, 2008, which is a divisional of U.S. application Ser. No. 10/244,295, filed Sep. 16, 2002, both of which are incorporated by reference herein in their entirety.

This application is generally related to U.S. application Ser. No. 09/659,214 filed Sep. 11, 2000 (now U.S. Pat. No. 6,760,457), which is hereby incorporated by reference.

This application is generally related to U.S. application Ser. No. 10/243,412 filed Sep. 12, 2002, which is hereby incorporated by reference.

FIELD OF THE INVENTION

This invention relates generally to hearing aids, and more particularly to switching structures and systems for a hearing aid.

BACKGROUND

Hearing aids can provide adjustable operational modes or characteristics that improve the performance of the hearing aid for a specific person or in a specific environment. Some of the operational characteristics are volume control, tone control, and selective signal input. One way to control these characteristics is by a manually engagable switch on the hearing aid. The hearing aid may include both a non-directional microphone and a directional microphone in a single hearing aid. Thus, when a person is talking to someone in a crowded room the hearing aid can be switched to the directional microphone in an attempt to directionally focus the reception of the hearing aid and prevent amplification of unwanted sounds from the surrounding environment. However, a conventional switch on the hearing aid is a switch that must be operated by hand. It can be a drawback to require manual or mechanical operation of a switch to change the input or operational characteristics of a hearing aid. Moreover, manually engaging a switch in a hearing aid that is mounted within the ear canal is difficult, and may be impossible, for people with impaired finger dexterity.

In some known hearing aids, magnetically activated switches are controlled through the use of magnetic actuators. For examples, see U.S. Pat. Nos. 5,553,152 and 5,659,621. The magnetic actuator is held adjacent the hearing aid and the magnetic switch changes the volume. However, such a hearing aid requires that a person have the magnetic actuator available when it desired to change the volume. Consequently, a person must carry an additional piece of equipment to control his\her hearing aid. Moreover, there are instances where a person may not have the magnetic actuator immediately present, for example, when in the yard or around the house.

Once the actuator is located and placed adjacent the hearing aid, this type of circuitry for changing the volume must cycle through the volume to arrive at the desired setting. Such an action takes time and adequate time may not be available to cycle through the settings to arrive at the required setting, for example, there may be insufficient time to arrive at the required volume when answering a telephone.

Some hearing aids have an input which receives the electromagnetic voice signal directly from the voice coil of a telephone instead of receiving the acoustic signal emanating

from the telephone speaker. Accordingly, signal conversion steps, namely, from electromagnetic to acoustic and acoustic back to electromagnetic, are removed and a higher quality voice signal reproduction may be transmitted to the person wearing the hearing aid. It may be desirable to quickly switch the hearing aid from a microphone (acoustic) input to a coil (electromagnetic field) input when answering and talking on a telephone. However, quickly manually switching the input of the hearing aid from a microphone to a voice coil, by a manual mechanical switch or by a magnetic actuator, may be difficult for some hearing aid wearers.

BRIEF DESCRIPTION OF THE DRAWINGS

A more complete understanding of the invention and its various features, objects and advantages may be obtained from a consideration of the following detailed description, the appended claims, and the attached drawings in which:

FIG. **1** illustrates the hearing aid of the present invention adjacent a magnetic field source;

FIG. **2** is a schematic view of the FIG. **1** hearing aid;

FIG. **3** shows a diagram of the switching circuit of FIG. **2**;

FIG. **4** is a schematic view of a hearing aid according to an embodiment of the present invention;

FIG. **5** is a schematic view of a hearing aid according to an embodiment of the present invention;

FIG. **6** is a schematic view of a hearing aid according to an embodiment of the present invention;

FIG. **7** is a schematic view of a hearing aid according to an embodiment of the present invention;

FIG. **8** is a schematic view of a hearing aid according to an embodiment of the present invention;

FIG. **9** is a schematic view of a hearing aid according to an embodiment of the present invention;

FIG. **10** is a schematic view of an embodiment of the present invention;

FIG. **11** is a circuit diagram of a power source of an embodiment of the present invention;

FIG. **12** is a circuit diagram of an embodiment of the present invention;

FIG. **13** is a circuit diagram of an embodiment of the present invention;

FIG. **14** is a schematic view of a hearing aid cleaning and charging system according to an embodiment of the present invention; and

FIG. **15** is a view of hearing aid switch of the present invention and a comparator/indicator circuit.

FIG. **16** is a diagram of a switching circuit according to an embodiment of the present invention.

FIG. **17** is a diagram of a switching circuit according to an embodiment of the present invention.

FIG. **18** is a diagram of a switching circuit according to an embodiment of the present invention.

FIG. **19** is a diagram of a switching circuit according to an embodiment of the present invention.

FIG. **20** is a diagram of a switching circuit according to an embodiment of the present invention.

FIG. **21** is a diagram of a switching circuit according to an embodiment of the present invention.

FIG. **22** is a diagram of a switching circuit according to an embodiment of the present invention.

DETAILED DESCRIPTION

In the following detailed description, reference is made to the accompanying drawings which form a part hereof and in which are shown by way of illustration specific embodiments in which the invention can be practiced. These embodiments are described in sufficient detail to enable those skilled in the art to practice and use the invention, and it is to be understood that other embodiments may be utilized and that electrical, logical, and structural changes may be made without departing from the spirit and scope of the present invention. The following detailed description is, therefore, not to be taken in a limiting sense and the scope of the present invention is defined by the appended claims and their equivalents.

Hearing aids provide different hearing assistance functions including, but not limited to, directional and non-directional inputs, multi-source inputs, filtering and multiple output settings. Hearing aids are also provide user specific and/or left or right ear specific functions such as frequency response, volume, varying inputs and signal processing. Accordingly, a hearing aid is programmable with respect to these functions or switch between functions based on the operating environment and the user's hearing assistance needs. A hearing aid is described that includes magnetically operated switches and programming structures.

One embodiment of the present invention provides a hearing aid that includes an input system, an output system, a signal processing circuit electrically connecting the input system to the output system, a magnetically actuatable switch between the input system and the signal processing circuit, and a filter connected to and controlled by the magnetically-actuatable switch. The switch allows the filter to filter a signal from the input system to the signal processing circuit or prevents the filter from filtering the signal. In an embodiment, the switch is a solid state switch. In an embodiment, the solid state switch is a giant magneto resistive (GMR) switch. In an embodiment, the solid state switch is an anisotropic magneto resistive (AMR) switch. In an embodiment, the solid state switch is a magnetic field effect transistor.

In an embodiment of the present invention, a magnetically actuatable switch is positioned between the output system and the signal processing circuit. This switch controls operation of a device before the output system or at the output system. In an embodiment, the switch selectively connects an output filter that filters the signal received by the output system. In an embodiment, the hearing aid includes a plurality of filters that are selectable based on the magnetic field sensed by the magnet switch or a magnetic field sensor.

An embodiment of the present invention provides a hearing aid that includes an input system, an output system, a programmable, signal processing circuit electrically connecting the input system to the output system, a magnetic field sensor, and a selection circuit connected to the magnetic sensor and at least one of the input system, output system and the signal processing system. The selection circuit is adapted to control the at least one of the input system, output system and the signal processing system based on a signal produced by the magnetic field sensor. The selection circuit is adapted to receive an electrical signal from the magnetic sensor and supply a programming signal to the signal processing circuit. In an embodiment, the magnetic field sensor is a full bridge circuit. In an embodiment, the magnetic field sensor is adapted to receive a pulsed power supply. In an embodiment, the selection circuit is connected to the input system and sends a control signal to the input system based on a signal received from the magnetic field sensor. In an embodiment, the input system includes a first input and a second input, and the input system activates one of the first input and the second input based on the control signal. The first input includes a microphone. The second input includes a magnetic field sensing device. The hearing aid of the present invention further includes a threshold circuit that blocks signals below a threshold value.

An embodiment of the present invention provides a hearing aid that includes a programming system that is adapted to sense a magnetic field and based on the magnetic field produce a programming signal. The programming signal, in an embodiment, includes a control sequence or code that allows the hearing aid to be programmed. The programming signal further includes a digital programming signal based on the magnetic field sensed by a magnetic field sensor.

An embodiment of the present invention includes a wireless on/off switch. The wireless on/off switch includes a magnetically operable switch. In an embodiment, the magnetically operable switch is a solid state switch. The on/off switch turns off the non-essential power to the hearing aid circuits to preserve battery power. In an embodiment, a system is provided that stores the hearing aid and provides a signal to turn off the hearing aid.

An embodiment of the invention includes a wireless switch that activates a power induction circuit in the hearing aid. The power induction circuit is adapted to receive a recharging signal from a power source and recharge the hearing aid power source. In an embodiment, the wireless switch that activates the power induction circuit also turns off the non-essential power consuming circuits of the hearing aid.

An embodiment of the invention includes a system that has a magnetic field source. In an embodiment, the magnetic field source being adapted to program the hearing aid. In an embodiment, the magnetic field source is adapted to wirelessly turn off and turn on the hearing aid. The system includes a storage receptacle for the hearing aid. In an embodiment, the magnetic field source provides a power induction signal that is adapted to recharge the hearing aid power source.

FIG. **1** illustrates an in-the-ear hearing aid **10** that is positioned completely in the ear canal **12**. A telephone handset **14** is positioned adjacent the ear **16** and, more particularly, the speaker **18** of the handset is adjacent the pinna **19** of ear **16**. Speaker **18** includes an electromagnetic transducer **21** which includes a permanent magnet **22** and a voice coil **23** fixed to a speaker cone (not shown). Briefly, the voice coil **23** receives the time-varying component of the electrical voice signal and moves relative to the stationary magnet **22**. The speaker cone moves with coil **23** and creates an audio pressure wave ("acoustic signal"). It has been found that when a person wearing a hearing aid uses a telephone it is more efficient for the hearing aid **10** to pick up the voice signal from the magnetic field gradient produced by the voice coil **23** and not the acoustic signal produced by the speaker cone.

Hearing aid **10** has two inputs, a microphone **31** and a voice coil pickup **32** (FIG. **2**). The microphone **31** receives acoustic signals, converts them into electrical signals and transmits same to a signal processing circuit **34**. The signal processing circuit **34** provides various signal processing functions which can include noise reduction, amplification, and tone control. The signal processing circuit **34** outputs an electrical signal to an output speaker **36** which transmits audio into the wearer's ear. The voice coil pickup **32** is an electromagnetic transducer, which senses the magnetic field gradient produced by movement of the telephone voice coil **23** and in turn produces a corresponding electrical signal which is transmitted to the signal processing circuit **34**. Accordingly, use of the voice coil pickup **32** eliminates two of the signal conversions normally necessary when a conventional hearing aid is used with a telephone, namely, the telephone handset **14** producing an acoustic signal and the hearing aid microphone **31** converting the acoustic signal to an electrical signal. It is believed that the elimination of these signal conversions improves the sound quality that a user will hear from the hearing aid.

A switching circuit **40** is provided to switch the hearing aid input from the microphone **31**, the default state, to the voice coil pickup **32**, the magnetic field sensing state. It is desired to automatically switch the states of the hearing aid **10** when the telephone handset **14** is adjacent the hearing aid wearer's ear. Thereby, the need for the wearer to manually switch the input state of the hearing aid when answering a telephone call and after the call is ends. Finding and changing the state of the switch on a miniaturized hearing aid can be difficult especially when the wearer is under the time constraints of a ringing telephone or if the hearing aid is an in the ear type hearing aid.

The switching circuit **40** of the described embodiment changes state when in the presence of the telephone handset magnet **22**, which produces a constant magnetic field that switches the hearing aid input from the microphone **31** to the voice coil pickup **32**. As shown in FIG. **3**, the switching circuit **40** includes a microphone activating first switch **51**, here shown as a transistor that has its collector connected to the microphone ground, base connected to a hearing aid voltage source through a resistor **58**, and emitter connected to ground. Thus, the default state of hearing aid **10** is switch **58** being on and the microphone circuit being complete. A second switch **52** is also shown as a transistor that has its collector connected to the hearing aid voltage source through a resistor **59**, base connected to the hearing aid voltage source through resistor **58**, and emitter connected to ground. A voice coil activating third switch **53** is also shown as a transistor that has its collector connected to the voice pick up ground, base connected to the collector of switch **52** and through resistor **59** to the hearing aid voltage source, and emitter connected to ground. A magnetically activated fourth switch **55** has one contact connected to the base of first switch **51** and through resistor **58** to the hearing aid voltage source, and the other contact is connected to ground. Contacts of switch **55** are normally open.

In this default open state of switch **55**, switches **51** and **52** are conducting. Therefore, switch **51** completes the circuit connecting microphone **31** to the signal processing circuit **34**. Switch **52** connects resistor **59** to ground and draws the voltage away from the base of switch **53** so that switch **53** is open and not conducting. Accordingly, hearing aid **10** is operating with microphone **31** active and the voice coil pickup **32** inactive.

Switch **55** is closed in the presence of a magnetic field, particularly in the presence of the magnetic field produced by telephone handset magnet **22**. In one embodiment of the invention, switch **55** is a reed switch, for example a micro-miniature reed switch, type HSR-003 manufactured by Hermetic Switch, Inc. of Chickasha, Okla. In a further embodiment of the invention, the switch **55** is a solid state, wirelessly operable switch. In an embodiment, wirelessly refers to a magnetic signal. An embodiment of a magnetic signal operable switch is a MAGFET. The MAGFET is non-conducting in a magnetic field that is not strong enough to turn on the device and is conducting in a magnetic field of sufficient strength to turn on the MAGFET. In a further embodiment, switch **55** is a micro-electro-mechanical system (MEMS) switch. In a further embodiment, the switch **55** is a magneto resistive device that has a large resistance in the absence of a magnetic field and has a very small resistance in the presence of a magnetic field. When the telephone handset magnet **22** is close enough to the hearing aid wearer's ear, the magnetic field produced by magnet **22** changes the state of switch (e.g., closes) switch **55**. Consequently, the base of switch **51** and the base of switch **52** are now grounded. Switches **51** and **52** stop conducting and microphone ground is no longer grounded. That is, the microphone circuit is open. Now switch **52** no longer draws the current away from the base of switch **53** and same is energized by the hearing aid voltage source through resistor **59**. Switch **53** is now conducting. Switch **53** connects the voice pickup coil ground to ground and completes the circuit including the voice coil pickup **32** and signal processing circuit **34**. Accordingly, the switching circuit **40** activates either the microphone (default) input **31** or the voice coil (magnetic field selected) input **32** but not both inputs simultaneously.

In operation, switch **55** automatically closes and conducts when it is in the presence of the magnetic field produced by telephone handset magnet **22**. This eliminates the need for the hearing aid wearer to find the switch, manually change switch state, and then answer the telephone. The wearer can conveniently, merely pickup the telephone handset and place it by his\her ear whereby hearing aid **10** automatically switches from receiving microphone (acoustic) input to receiving pickup coil (electromagnetic) input. That is, a static electromagnetic field causes the hearing aid to switch from an audio input to a time-varying electro-magnetic field input. Additionally, hearing aid **10** automatically switches back to microphone input after the telephone handset **14** is removed from the ear. This is not only advantageous when the telephone conversation is complete but also when the wearer needs to talk with someone present (microphone input) and then return to talk with the person on the phone (voice coil input).

The above described embodiment of the switching circuit **40** describes a circuit that grounds an input and open circuits the other inputs. It will be recognized that the switching circuit **40**, in an embodiment, connects the power source to an input and disconnects the power source to the other inputs. For example, the collectors of the transistors **51** and **53** are connected to the power source. The switch **55** remains connected to ground. The emitter of transistor **51** is connected to the power input of the microphone **31**. The emitter of the transistor **53** is connected to the power input of the voice coil **32**. Thus, switching the switch **55** causes the power source to be interrupted to the microphone and supplied to the voice coil pickup **32**. In an embodiment, switching circuit **40** electrically connects the signal from one input to the processing circuit **34** and opens (disconnects) the other inputs from the processing circuit **34**.

While the disclosed embodiment references an in-the-ear hearing aid, it will be recognized that the inventive features of the present invention are adaptable to other styles of hearing aids including over-the-ear, behind-the-ear, eye glass mount, implants, body worn aids, etc. Due to the miniaturization of hearing aids, the present invention is advantageous to many miniaturized hearing aids.

FIG. **4** shows hearing aid **70**. The hearing aid **70** includes a switching circuit **40**, a signal processing circuit **34** and an output speaker **36** as described herein. The switching circuit **40** includes a magnetic field responsive, solid state circuit. The switching circuit **40** selects between a first input **71** and a second input **72**. In an embodiment, the first input **71** is an omnidirectional microphone, which detects acoustical signals in a broad pattern. In an embodiment, the second input **72** is a directional microphone, which detects acoustical signals in a narrow pattern. The omnidirectional, first input **71** is the default state of the hearing aid **70**. When the switching circuit **40** senses the magnetic field, the switch changes state from its default to a magnetic field sensed state. The magnetic field sensed state causes the hearing aid **70** to switch from its default mode and the directional, second input **72** is activated. In an embodiment, the activation of the second input **72** is mutually exclusive of activation of the first input **71**.

In use with a telephone handset, e.g., **14** shown in FIG. **1**, hearing aid **70** changes from its default state with omnidirectional input **71** active to its directional state with directional input **72** active. Thus, hearing aid **70** receives its input acoustically from the telephone handset. In an embodiment, the directional input **72** is tuned to receive signals from a telephone handset.

In an embodiment, switching circuit **40** includes a micro-electro-mechanical system (MEMS) switch. The MEMS switch includes a cantilevered arm that in a first position completes an electrical connection and in a second position opens the electrical connection. When used in the circuit as shown in FIG. **3**, the MEMS switch is used as switch **55** and has a normally open position. When in the presence of a magnetic field, the cantilevered arm shorts the power supply to ground. This initiates a change in the operating state of the hearing aid input.

FIG. **5** shows an embodiment of a hearing aid **80** according to the teachings of the present invention. Hearing aid **80** includes at least one input **81** connected to a signal processing circuit **34**, which is connected to an output speaker **36**. In an embodiment, hearing aid **80** includes two or more inputs **81** (one shown). The input **81** includes a signal receiver **83** that includes two nodes **84**, **85**. Node **84** is connected to the signal processing circuit **34** and to one terminal of a capacitor **86**. In an embodiment, node **84** is the negative terminal of the input **81**. In an embodiment, node **84** is the ground terminal of the input **81**. Node **85** is connected to one pole of a magnetically operable switch **87**. In an embodiment, the switch **87** is a mechanical switch, such as a reed switch. In an embodiment, the switch **87** is a solid-state, magnetically actuated switch circuit. In an embodiment, the switch **87** is a micro-electro-mechanical system (MEMS). In an embodiment, the solid state switch **87** is a MAGFET. In an embodiment, the solid state switch **87** is a giant magneto-resistivity (GMR) sensor. In an embodiment, the switch **87** is normally open. The other pole of switch **87** is connected to the second terminal of capacitor **86** and to the signal processing circuit **34**. Switch **87** automatically closes when in the presence of a magnetic field. When the switch **87** is closed, input **81** provides a signal that is filtered by capacitor **86**. The filtered signal is provided to the signal processing circuit **34**. The capacitor **86** acts as a filter for the signal sent by the input **81** to the signal processing circuit **34**. Thus, switch **87** automatically activates input **81** and filter **86** when in the presence of a magnetic (wireless) field or signal. When the magnetic field is removed, then the switch automatically opens and electrically opens the input **81** and filter **86** from the signal processing circuit **34**.

FIG. **6** shows a further hearing aid **90**. Hearing aid **90** includes at least one input **81** having nodes **84**, **85** connected to signal processing circuit **34**, which is connected to output speaker **36**. Node **85** is connected to first pole of switch **87**. Node **84** is connected to a first terminal of filter **86**. The second pole of switch **87** is connected to the second terminal of filter **86**. In an embodiment, the switch **87** is normally open. Accordingly, in the default state of hearing aid **90**, the signal sensed by input **81** is sent directly to the signal processing circuit **34**. In the switch active state of hearing aid **90**, the switch **87** is closed and the signal sent from the input **81** is filtered by filter **86** prior to the signal being received by the signal processing circuit **34**. The FIG. **6** embodiment provides automatic signal filtering when the switch **87**, and hence the hearing aid **90**, is in the presence of a magnetic field.

FIG. **7** shows a further hearing aid **100** that includes input **81**, signal processing circuit **34** and output system **36**. The input **81** is connected to a plurality of filtering circuits $\mathbf{101}_1$, $\mathbf{101}_2$, $\mathbf{101}_3$. Thus, signal generated by the input **81** is applied to each of the filters **101**. Each of the filtering circuits **101** provides a different filter effect. For example, the first filter is a low-pass filter. The second filter is a high-pass filter. The third filter is a low-pass filter. In an embodiment, at least one of filtering circuits $\mathbf{101}_1$, $\mathbf{101}_2$, $\mathbf{101}_3$ includes an active filter. Each of the filters **101** are connected to a switching circuit **102**. In an embodiment, the switching circuit **102** is a magnetically actuatable switch as described herein. The switching circuit **102** determines which of the filters **101** provides a filtered signal to the signal processing circuit **34**. The processing circuit **34** sends a signal to the output system **36** for broadcasting into the ear of the hearing aid wearer. The switching circuit **102** in the absence of a magnetic field electrically connects the first filter $\mathbf{101}_1$ to the signal processing circuit **34** and electrically opens the second filter $\mathbf{101}_2$ and third filter $\mathbf{101}_3$. The switching circuit **102** in the presence of a magnetic field opens the first filter $\mathbf{101}_1$ and electrically connects at least one of the second filter $\mathbf{101}_2$ and third filter $\mathbf{101}_3$ to the signal processing circuit **34**. In an embodiment, the second and third filters provide a band-pass filter with both being activated by the switching circuit **102**. While the embodiment of FIG. **7** shows the switching circuit **102** positioned between the filters and the hearing aid signal processing circuit **34**, the switching circuit **102** is positioned between the input **81** and the filtering circuits $\mathbf{101}_1$, $\mathbf{101}_2$, $\mathbf{101}_3$ in an embodiment of the present invention. In this embodiment, the switching circuit **102** only supplies the input signal from input **81** to the selected filtering circuit(s) $\mathbf{101}_1$, $\mathbf{101}_2$, $\mathbf{101}_3$.

FIG. **8** shows an embodiment of the present invention including a hearing aid **110** having a magnetic field sensor **115**. The magnetic field sensor **115** is connected to a selection circuit **118**. The selection circuit **118** controls operation of at least one of a programming circuit **120**, a signal processing circuit **122**, output processing circuit **124** and an input circuit **126**. The sensor **115** senses a magnetic field or signal and outputs a signal to the selection circuit **118**, which controls at least one of circuits **120**, **122**, **124** and **126** based on the signal produced by the magnetic field sensor **115**. The signal output by sensor **115** includes an amplitude level that may control which of the circuits that is selected by the selection circuit **118**. That is, a magnetic field having a first strength as sensed by sensor **115** controls the input **126**. A magnetic field having a second strength as sensed by sensor **115** controls the programming circuit **120**. The magnetic field as sensed by sensor **115** then varies from the second strength to produce a digital programming signal. In an embodiment, the signal output by sensor **115** includes digital data that is interpreted by the selection circuit to select at least one of the subsequent circuits. The selection circuit **118** further provides a signal to the at least one of the subsequent circuits. The signal controls operation of the at least one circuit.

In an embodiment, the signal from the selection circuit **118** controls operation of a programming circuit **120**. Programming circuit **120** provides hearing aid programmable settings to the signal processing circuit **122**. In an embodiment, the magnetic sensor **115** and the selection circuit **118** produce a digital programming signal that is received by the programming circuit **120**. Hearing aid **110** is programmed to an individual's specific hearing assistance needs by providing programmable settings or parameters to the hearing aid. Programmable settings or parameters in hearing aids include, but are not limited to, at least one of stored program selection, frequency response, volume, gain, filtering, limiting, and attenuation. The programming circuit **120** programs the programmable parameters for the signal processing circuit **122** of the hearing aid **110** in response to the programming signal received from the magnetic sensor **115** and sent to the programming circuit **120** through selection circuit **118**.

In an embodiment, the signal from selection circuit **118** directly controls operation of the signal processing circuit **122**. The signal received by the processing circuit **122** controls at least one of the programmable parameters. Thus, while the signal is sent by the magnetic sensor **115** and the selection circuit **118**, the programmable parameter of the signal processing circuit **122** is altered from its programmed setting based on the signal sensed by the magnetic field sensor **115** and sent to the signal processing circuit **122** by the selection circuit **118**. It will be appreciated that the programmed setting is a factory default setting or a setting programmed for an individual. In an embodiment, the alteration of the hearing aid settings occurs only while the magnetic sensor **115** senses the magnetic field. The hearing aid **110** returns to its programmed settings after the magnetic sensor **115** no longer senses the magnetic field.

In an embodiment, the signal from selection circuit **118** directly controls operation of the output processing circuit **124**. The output processing circuit **124** receives the processed signal, which represents a conditioned audio signal to be broadcast into a hearing aid wearer's ear, from the signal processing circuit **122** and outputs a signal to the output **128**. The output **128** includes a speaker that broadcasts an audio signal into the user's ear. Output processing circuit **124** includes filters for limiting the frequency range of the signal broadcast from the output **128**. The output processing circuit **124** further includes an amplifier for amplifying the signal between the signal processing circuit **122** and the output. Amplifying the signal at the output allows signal processing to be performed at a lower power. The selection circuit **118** sends a control signal to the output processing circuit **124** to control the operation of at least one of the amplifying or the filtering of the output processing circuit **124**. In an embodiment, the output processing circuit **124** returns to its programmed state after the magnetic sensor **115** no longer senses a magnetic field.

In an embodiment, the signal from the selection circuit **118** controls operation of the input circuit **126** to control which input is used. For example, the input circuit **126** includes a plurality of inputs, e.g., an audio microphone and a magnetic field input or includes two audio inputs. In an embodiment, the input circuit **126** includes an omnidirectional microphone and a directional microphone. The signal from the selection circuit **118** controls which of these inputs of the input circuit **126** is selected. The selected input sends a sensed input signal, which represents an audio signal to be presented to the hearing aid wearer, to the signal processing circuit **122**. In a further example, the input circuit **126** includes a filter circuit that is activated and/or selected by the signal produced by the selection circuit **118**.

FIG. **9** shows an embodiment of the magnetic sensor **115**. Sensor **115** includes a full bridge **140** that has first node connected to power supply (Vs) and a second node connected ground. The bridge **140** includes third and fourth nodes whereat the sensed signal is output to further hearing aid circuitry. A first variable resistor R**1** is connected between the voltage source and the third node. A second variable resistor R**2** is connected between ground and the fourth node. The first and second variable resistors R**1** and R**2** are both variable based on a wireless signal. In an embodiment, the wireless signal includes a magnetic field signal. A first fixed value resistor R**3** is connected between the voltage source and the fourth node. A second fixed value resistor R**4** is connected between ground and the third node. The bridge **140** senses an electromagnetic field produced by a source **142** and produces a signal that is fed to an amplifier **143**. Both the first and second variable resistors R**1** and R**2** vary in response to the magnetic field produced by magnetic field source **142**. Amplifier **143** amplifies the sensed signal. A low pass filter **144** filters high frequency components from the signal output by the amplifier **143**. A threshold adjust circuit **145**, which is controlled by threshold control circuit **146**, adjusts the level of the signal prior to supplying it to the selection circuit **118**. In an embodiment, the threshold adjust circuit **145** holds the level of the signal below a maximum level. The maximum level is set by the threshold adjust circuit **146**.

FIG. **10** shows a further embodiment of magnetic sensor **115**, which includes a half bridge **150**. The half bridge **150** includes two fixed resistors R**5**, R**6** connected in series between a voltage source and the output node. Bridge **150** further includes two variable resistors R**7**, R**8** connected in series between ground and the output node. The two variable resistors R**7**, R**8** sense the electromagnetic field produced by the magnetic field source **142** to produce a corresponding signal at the output node. The amplifier **143**, filter **144**, threshold adjust circuit **145** and selection circuit **118** are similar to the circuits described herein.

The magnetic sensor **115**, in either the full bridge **140** or half bridge **150**, includes a wireless signal responsive, solid state device. The solid state sensor **115**, in an embodiment, includes a giant magnetoresistivity (GMR) device, which relies on the changing resistance of materials in the presence of a magnetic field. One such GMR sensor is marketed by NVE Corp. of Eden Prairie, Minn. under part no. AA002-02. In one embodiment of a GMR device, a plurality of layers are formed on a substrate or wafer to form an integrated circuit device. Integrated circuit devices are desirable in hearing aids due to their small size and low power consumption. A first layer has a fixed direction of magnetization. A second layer has a variable direction of magnetization that depends on the magnetic field in which it is immersed. A non-magnetic, conductive layer separates the first and second magnetic layers. When the direction of magnetization of the first and second layers are the same, the resistance across the GMR device layer is low. When the direction of magnetization of the second layer is at an angle with respect to the first layer, then the resistance across in the layers increases. Typically, the maximum resistance is achieved when the direction of magnetization are at an angle of about 180 degrees. Such GMR devices are manufactured using VLSI fabrication techniques. This results in magnetic field sensors having a small size, which is also desirable in hearing aids. In an embodiment, a GMR sensor of the present invention has an area of about 130 mil by 17 mil. It will be appreciated that smaller GMR sensors are desirable for use in hearing aids if they have the required sensitivity and bandwidth. Further, some hearing aids are manufactured on a ceramic substrate that will form a base layer on which a GMR sensor is fabricated. GMR sensors have a low sensitivity and thus must be in a strong magnetic field to sense changes in the magnetic field. Further, magnetic field strength depends on the cube of the distance from the source. Accordingly, when the GMR sensor is used to program a hearing aid, the magnetic field source **142** must be close to the GMR sensor. As a example, a programming coil of the source **142** is positioned about 0.5 cm from the GMR sensor to provide a strong magnetic field to be sensed by the magnetic field sensor **115**.

When the GMR sensor is used in the hearing aid circuits described herein, the GMR sensor acts as a switch when it senses a magnetic field having at least a minimum strength. The GMR sensor is adapted to provide various switching functions. The GMR sensor acts as a telecoil switch when it is placed in the DC magnetic field of a telephone handset in a first function. The GMR sensor acts as a filter-selecting switch that electrically activates or electrically removes a filter from the signal processing circuits of a hearing aid in an embodiment. The GMR sensor acts to switch the hearing aid input in an embodiment. For example, the hearing aid switches between acoustic input and magnetic field input. As a further example, the hearing aid switches between omni-directional input and directional input. In an embodiment, the GMR sensor acts to automatically turn the power off when a magnetic field of sufficient strength changes the state, i.e., increases the resistance, of the GMR sensor.

The GMR sensor is adapted to be used in a hearing aid to provide a programming signal. The GMR sensor has a bandwidth of at least 1 MHz. Accordingly, the GMR sensor has a high data rate that is used to program the hearing aid during manufacture. The programming signal is a digital signal produced by the state of the GMR sensor when an alternating or changing magnetic field is applied to the GMR sensor. For example, the magnetic field alternates about a threshold field strength. The GMR sensor changes its resistance based on the magnetic field. The hearing aid circuit senses the change in resistance and produces a digital (high or low) signal based on the GMR sensor resistance. In a further embodiment, the GMR sensor is a switch that activates a programming circuit in the hearing aid. The programming circuit in an embodiment receives audio signals that program the hearing aid. In an embodiment, the audio programming signal is broadcast through a telephone network to the hearing aid. Thus, the hearing aid is remotely programmed over a telephone network using audio signals by non-manually switching the hearing aid to a programming mode. In an embodiment, the hearing aid receives a variable magnetic signal that programs the hearing aid. In an embodiment, the telephone handset produces the magnetic signal. The continuous magnetic signal causes the hearing aid to switch on the programming circuit. The magnetic field will remain above a programming threshold. The magnetic field varies above the programming threshold to produce the programming signal that is sensed by the magnetic sensor and programs the hearing aid. In a further embodiment, a hearing aid programmer is the source of the programming signal.

The solid state sensor **115**, in an embodiment, is an anisotropic magneto resistivity (AMR) device. An AMR device includes a material that changes its electrical conductivity based on the magnetic field sensed by the device. An example of an AMR device includes a layer of ferrite magnetic material. An example of an AMR device includes a crystalline material layer. In an embodiment, the crystalline layer is an orthorhombic compound. The orthorhombic compound includes RCu2 where R=a rare earth element). Other types of anisotropic materials include anisotropic strontium and anisotropic barium. The AMR device is adapted to act as a hearing aid switch as described herein. That is, the AMR device changes its conductivity based on a sensed magnetic field to switch on or off elements or circuits in the hearing aid. The AMR device, in an embodiment, is adapted to act as a hearing aid programming device as described herein. The AMR device senses the change in the state of the magnetic field to produce a digital programming signal in the hearing aid.

The solid state sensor **115**, in an embodiment, is a spin dependent tunneling (SDT) device. Spin dependent tunneling (SDT) structures include an extremely thin insulating layer separating two magnetic layers. The conduction is due to quantum tunneling through the insulator. The size of the tunneling current between the two magnetic layers is modulated by the magnetization directions in the magnetic layers. The conduction path must be perpendicular to the plane of a GMR material layer since there is such a large difference between the conductivity of the tunneling path and that of any path in the plane. Extremely small SDT devices with high resistance are fabricated using photolithography allowing very dense packing of magnetic sensors in small areas. The saturation fields depend upon the composition of the magnetic layers and the method of achieving parallel and antiparallel alignment. Values of a saturation field range from 0.1 to 10 kA/m (1 to 100 Oe) offering the possibility of extremely sensitive magnetic sensors with very high resistance suitable for use with battery powered devices such as hearing aids. The SDT device is adapted to be used as a hearing aid switch as described herein. The SDT device is further adapted to provide a hearing aid programming signals as described herein.

Hearing aids are powered by batteries. In an embodiment, the battery provides about 1.25 Volts. A magnetic sensor, e.g., bridges **140** or **150**, sets the resistors at 5K ohms, with the variable resistors R**1**, R**2** or R**7**, R**8** varying from the 5K ohm dependent on the magnetic field. In this embodiment, the magnetic sensor **140** or **150** would continuously draw about 250 μA. It is desirable to limit the power draw from the battery to prolong the battery life. One construction for limiting the power drawn by the sensor **140** or **150** is to pulse the supply voltage Vs. FIG. **11** shows a pulsed power circuit **180** that receives the 1.25 Volt supply from the hearing aid battery **181**. Pulsed power circuit **180** includes a timer circuit that is biased (using resistors and capacitors) to produce a 40 Hz pulsed signal that has a pulse width of about 2.8 μsec. and a period of about 25.6 μsec for a duty cycle of about 0.109. Such, a pulsed power supply uses only about a tenth of the current that a continuous power supply would require. Thus, with a GMR sensor that continuously draws 250 μA, would only draw about 25 μA to with a pulsed power supply. In the specific embodiment, the current drain on the battery would be about 27 μA to (0.109*250 μA). Accordingly, the power savings of a pulsed power supply versus a continuous power supply is about 89.1%.

FIG. **12** shows an embodiment of a GMR sensor circuit **190** that operates as both a hearing aid state changing switch and as a programming circuit. Circuit **190** includes a sensing stage **192**, followed by a high frequency signal stage **193**, which is followed by a bi-state sensing and switch stage **201**. The hearing aid state changing switch is adaptable to provide any of bi-states of the hearing aid, for example, changing inputs, changing filters, turning the hearing aid on or off, etc. The GMR sensor circuit **190** includes a full bridge **192** that receives a source voltage, for example, Vs or the output from the pulse circuit **180**. Vs is, in an embodiment, the battery power. The bridge **192** outputs a signal to both the signal stage **193** and the switch stage **201**. The positive and negative output nodes of the full bridge **192** are respectively connected to the non-inverting and inverting terminals of an amplifier **194** through capacitors **195**, **196**. The amplifier is part of the signal stage **193**. In an embodiment, the output **197** of the amplifier **194** is a digital signal that is used to program the hearing aid. The hearing aid programming circuit, e.g., programming circuit **120**, receives the digital signal **197** from the amplifier **194**. The signal **197**, in an embodiment, is the audio signal that is inductively sensed by bridge **192** and is used as an input to the hearing aid signal processing circuit.

The switching stage **201** includes filters to remove the high frequency component of the signal from the induction sensor. The positive and negative output nodes of the full bridge **192** are each connected to a filter **198**, **199**. Each filter **198**, **199** includes a large resistor (1 M ohm) and a large capacitor (1 μf). The filters **198**, **199** act to block false triggering of the on/off switch component **200** of the circuit **190**. The signals that pass filters **198**, **199** are fed through a series of amplifiers to determine whether an electromagnetic field is present to switch the state of the hearing aid. An output **205** is the on/off signal from the on/off switch component **200**. The on/off signal is used to select one of two states of the hearing aid. The state of the hearing aid, in an embodiment, is between an audio or electromagnetic field input. In another embodiment, the state of the hearing aid is either an omni-directional input or directional input. In an embodiment, the state of the hearing aid is a filter acting on a signal in the hearing aid or not. In an embodiment, the signal **205** is sent to a level detection circuit **206**. Level detection circuit **206** outputs a digital (high or low) signal **207** based on the level of signal **205**. In this embodiment, signal **207** is the signal used for switching the state of the hearing aid.

FIG. **13** shows a saturated core circuit **1300** for a hearing aid. The saturated core circuit **1300** senses a magnetic field and operates a switch or provides a digital programming signal. A pulse circuit **1305** connects the saturated core circuit to the power supply Vs. Pulse circuit **1305** reduces the power consumption of the saturated core circuit **1300** to preserve battery life in the hearing aid. The pulse circuit **1305** in the illustrated embodiment outputs a 1 MHz signal, which is fed to a saturatable core, magnetic field sensing device **1307**. In an embodiment, the device includes a magnetic field sensitive core wrapped by a fine wire. The core in an example is a 3.0×0.3 mm core. In an embodiment, the core is smaller than 3.0×0.3 mm. The smaller the core, the faster it responds to magnet fields and will saturate faster with a less intense magnetic field. An example of a saturated core is a telecoil marketed by Tibbetts Industries, Inc. of Camden, Me. However, the present invention is not limited to the Tibbetts Industries telecoil. In a preferred embodiment of the invention, the saturatable core device **1307** is significantly smaller than a telecoil so that the device will saturate faster in the presence of the magnetic field. The device **1307** changes in A.C. impedance based on the magnetic field surrounding the core. The core has a first impedance in the presence of a strong magnetic field and a second impedance when outside the presence of a magnetic field. A resistor **1308** connects the device **1307** to ground. In an embodiment, the resistor **1308** has a value of 100 KOhms. The node intermediate the device **1307** and resistor **1308** is a sensed signal output that is based on the change in impedance of the device **1307**. Accordingly, the saturable core device **1307** and resistor **1308** act as a half bridge or voltage divider. The electrical signal produced by the magnetic field sensing device **1307** and resistor **1308** is sent through a diode D1 to rectify the signal. A filter **1309** filters the rectified signal and supplies the filtered signal to an input of a comparator **1310**. The comparator **1310** compares the signal produced by the filter and magnetic field sensor to a reference signal to produce output signal **1312**. In an embodiment, the signal output through the core device **1307** varies +/−40 mV depending on the magnetic field in which the saturable core device **1307** is placed. In an embodiment, it is preferred that the magnetic field is of sufficient strength to move the saturable core device into saturation. While device **1307** is shown as a passive device, in an embodiment of the present invention, device **1307** is a powered device. In an embodiment, the saturatable device **1307** acts a non-manual switch that activates or removes circuits from the hearing aid circuit. For example, the saturatable device **1307** acts to change the input of the hearing aid in an embodiment. In a further embodiment, the saturated core circuit **1300** activates or removes a filter from the hearing aid circuit based on the state of the output **1312**. In a further embodiment, the saturatable core device **1307** is adapted to be a telecoil switch. In a further embodiment, the saturatable core device **1307** is adapted to act as a automatic, non-manual power on/off switch. In a further embodiment, the saturatable core **1307** is a programming signal receiver.

FIG. **14** shows a system **1401** including a hearing aid **1405** and a hearing aid storage receptacle **1410**. Receptacle **1410** is cup-like with an open top **1411**, an encircling sidewall **1412** upstanding from a base **1413**. The receptacle **1410** is adapted to receive the hearing aid **1405** and store it adjacent a magnetic field source **1415**. The receptacle base **1413** houses the magnetic field source **1415**. Thus, when the hearing aid **1405** is in the receptacle (shown in solid line in FIG. **14**), the hearing aid is in the magnetic field. In an embodiment, the magnetic field experienced by the hearing aid in the receptacle is the near field. When the hearing aid **1405** is out of receptacle (broken line showing in FIG. **14**), the hearing aid is out of the magnetic field, i.e., the magnetic field does not have sufficient strength as sensed by the magnetic field sensor of hearing aid **1405** to trigger a state changing signal in the hearing aid **1405**. In an embodiment, the hearing aid **1405** includes a magnetically-actuated switch **1406**. The magnetically-actuated switch **1406** is a normally on (conducting) switch that connects the power supply to the hearing aid circuit. When the hearing aid **1405** is in the receptacle, the magnetically-actuated switch changes to a non-conducting state and the power supply is electrically disconnected from the hearing aid circuit. Thus, hearing aid **1405** is placed in a stand-by mode. The stand-by mode reduces power consumption by the hearing aid. This extends hearing aid battery life. Moreover, this embodiment eliminates the need for the hearing aid wearer to manually turn off the hearing aid after removing it. The wearer merely places the hearing aid **1405** in the storage receptacle **1410** and the hearing aid **1405** turns off or is placed in a stand-by mode. Non-essential power draining circuits are turned off. Non-essential circuits include those that are used for signal processing that are not needed when the hearing aid wearer removes the hearing aid. The stand-by mode is used so that any programmable parameters stored in the hearing aid **1405** are saved in memory by power supplied to the hearing aid memory. The programmable parameters are essential parameters that are stored in the hearing aid and should not be deleted with the power being turned off. The programmed parameters include the volume level. Thus, when the hearing aid **1405** is removed from the receptacle **1410**, the hearing aid is automatically powered by the normally on switch **1406** electrically reconnecting the hearing aid signal processing circuit to the power supply and the hearing aid **1405** returns to the stored volume level without the wearer being forced to manually adjust the volume level of the hearing aid.

The hearing aid storage system **1401**, in an embodiment, includes a magnetic field source **1415** that produces a magnetic field that is significantly greater, e.g., at least 3-4 times as great, as the constant magnetic field and/or the varying magnetic field of a telephone handset. This allows the hearing aid **1405** to include both the automatic switch **40** that alternates inputs based on a magnetic field of a first threshold and the automatic power-off switch **1406** that turns off the hearing aid based on a magnetic field of a higher threshold. Thus, hearing aid **1405** includes automatically switching between inputs, filters, settings, etc. as described herein and automatically powering down to preserve battery power when the hearing aid is in the storage receptacle **1410**.

In another embodiment of the present invention, the hearing aid **1405** further includes a rechargeable power supply **1407** and a magnetically actuated switching circuit **1406** as described herein. The rechargeable power supply **1407** includes at least one of a rechargeable battery. In an embodiment, rechargeable power supply **1407** includes a capacitor. In an embodiment, a power induction receiver is connected to the rechargeable power supply **1407** through the switching circuit **1406**. The receptacle **1410** includes a power induction transmitter **1417** and magnetic field source **1415**. When the hearing aid **1405** is positioned in the receptacle **1410**, the magnetic switch **1406** turns on a power induction receiver of the rechargeable power supply **1407**. The power induction receiver receives a power signal from the power induction transmitter **1417** to charge the power supply **1407**. Thus, whenever the hearing aid **1405** is stored in the receptacle **1410**, the hearing aid power supply **1407** is recharged. In an embodiment, the magnetically actuated switch **1406** electrically disconnects the hearing aid circuit from the hearing aid power supply **1407** and activates the power induction receiver to charge the hearing aid power supply. As a result, the hearing aid power supply **1407** is recharged when the hearing aid is not in use by the wearer.

In a further embodiment, the system **1401** includes a cleaning source **1430** connected to the storage receptacle **1410**. The cleaning source **1430** supplies sonic or ultrasonic cleaning waves inside the receptacle **1411**. The waves are adapted to clean the hearing aid **1405**. Accordingly, the hearing aid **1405** is automatically cleaned when placed in the receptacle **1411**.

FIG. **15** shows a further embodiment of the hearing aid switch **1406** that includes an indicator circuit **1450**. Indicator circuit **1450** is adapted to produce an indicator signal to the hearing aid user. In an embodiment, the indicator circuit **1450** is connected to a magnetic field sensor, e.g. sensor **115**, **190** or **1300**. The indicator circuit provides an indication signal that indicates that the magnetic field sensor **190** or **1300** is sensing the magnetic field. In an embodiment, the indicator circuit indicates that the hearing aid has been disconnected from the power supply. In an embodiment, the indicator circuit indicates that the hearing aid power supply is being recharged by the recharging circuit **1417**. Indicator circuit **1450** includes a comparator **1455** that receives the output signal from the magnetic field sensor circuit **190** or **1300** and compares the received output signal to a threshold value and based on the comparison sends a signal to an indicator **1460** that produces the indicator signal. The indicator signal is a visual signal produced by a low power LED.

FIG. **16** shows a hearing aid switch circuit **1600**. Circuit **1600** switches the power from one input to another input. In an embodiment, one input is an induction input and the other input is an audio input. In an embodiment, circuit **1600** exclusively powers one of the inputs. Circuit **1600** includes a power supply **1601** connected to a resistor **1603** at node **1604**. Hence, node **1604** is at a high, non-groung potential. In an embodiment, the power supply is a hearing aid battery power supply. In an embodiment, the power supply is in the range of 1.5 to 0.9 volts. In an embodiment, the resistor **1603** is a 100 KOhm. The resistor **1603** is connected to a non-manual switch **1605** that is connected to ground. Switch **1605**, in an embodiment, is a magnetically actuatable switch as described herein. An input to first invertor **1607** is connected to node **1604**. The output of invertor **1607** is connected to the input of a first hearing aid input **1609** and an input of a second invertor **1611**. The output of the second invertor **1611** is connected to a second hearing aid input **1613**. In an embodiment, first and second invertors **1607** and **1611** are Fairchild ULP-A NC7SV04 invertors. The invertors have an input voltage range from 0.9V to 3.6V.

The circuit **1600** has two states. In the first state, which is illustrated, the switch **1605** is open. The node **1604** is at a high voltage. Invertor **1607** outputs a low signal, which is supplied to both the first input **1609** and the second invertor **1611**. The first input **1609** is off when it receives a low signal. The second invertor **1611** outputs a high, on signal to the second input **1613**. Accordingly, in the open switch state of circuit **1600**, the first input **1609** is off and the second input **1613** is on. When in the presence of a magnetic field, switch **1605** closes. Node **1604** is connected to ground and, hence, is at a low potential. Invertor **1607** outputs a high, on signal to the first input **1609** and second invertor **1611**. The first input **1609** is on, i.e., powered. The second invertor **1611** outputs a low, off signal to second input **1613**. Accordingly, in the closed switch state of circuit **1600**, the first input **1609** is on and the second input **1613** is off. In an embodiment, the first hearing aid input **1609** is an induction input and the second hearing aid input **1613** is an audio input. Thus, in the switch open state, the second, audio input **1613** is on or powered and the first, induction input **1609** is off or unpowered. In the switch closed state, the first, induction input **1609** is on or powered and the second, audio input **1613** is off. The circuit **1600** is used as an automatic, induction telephone signal input circuit.

FIG. **17** shows a hearing aid switch circuit **1700**. Circuit **1700** is similar to circuit **1600**, like elements are designated with the same two least significant digits and the two most significant digit refer to the FIG. on which they appear. In circuit **1700**, the switch **1705** is connected to the voltage supply **1701**. Resistor **1703** is connected between node **1704** and ground. The input of first invertor **1707** is connected to node **1704**. The output of first invertor **1707** is connected to the first input **1709** and the input of the second invertor **1711**. The output of the second invertor **1711** is connected to the second input **1713**.

The circuit **1700** has two states. In the first state, which is illustrated, the switch **1705** is open. The node **1704** is grounded by resistor **1703** and is at a low potential. Invertor **1707** outputs a high signal, which is supplied to both the first input **1709** and the second invertor **1711**. The first input **1709** is on when it receives a high signal. The second invertor **1711** outputs a low, off signal to the second input **1713**. Accordingly, in the open switch state of circuit **1700**, the first input **1709** is on and the second input **1713** is off. When in the presence of a magnetic field, switch **1705** closes. Node **1704** is connected to the voltage supply through closed switch **1705** and, hence, is at a high potential. Invertor **1707** outputs a low, off signal to the first input **1709** and second invertor **1711**. The first input **1709** is off, i.e., unpowered. The second invertor **1711** outputs a high, on signal to second input **1713**. Accordingly, in the closed switch state of circuit **1700**, the first input **1709** is off and the second input **1713** is on. In an embodiment, the first hearing aid input **1709** is an audio input and the second hearing aid input **1713** is an induction input. Thus, in the switch open state, the first, audio input **1709** is on or powered and the second, induction input **1713** is off or unpowered. In the switch closed state, the first, audio input **1709** is off and the second, induction input **1713** is on or powered. The circuit **1700** is used as an automatic, induction telephone signal input circuit. Further, circuit **1700** does not continually incur the loss associated with resistor **1703**. The default state of the circuit **1700** is with the resistor **1703** grounded and no power drain occurs across resistor **1703**. In circuit **1600**, there is a continuous power loss associated with resistor **1603**. Power conservation and judicious use of the battery power in a hearing aid is a significant design characteristic.

FIG. **18** shows a hearing aid switch circuit **1800**. Circuit **1800** includes a supply voltage **1801** connected to an induction, first hearing aid input **1809** and a non-manual switch **1805**. Switch **1805**, in an embodiment, is a magnetic field actuatable switch as described herein. A resistor **1803** connects a node **1804** to ground. Switch **1805** is connected to node **1804**. Invertor **1807** is connected to node **1810**. Both first input **1809** and an audio, second hearing aid input **1813** are connected to node **1810**. Second input **1813** is connected to ground. Circuit **1800** has two states. In a first, switch open state node **1804** is connected to ground through resistor **1803**. The invertor **1807** outputs a high signal to node **1810**. The high signal turns on or powers the second input **1813**. The high signal at node **1810** is a high enough voltage to hold the potential across the first input **1809** to be essential zero. In an embodiment, the high signal output by invertor **1807** is essentially equal to the supply voltage **1801**. Thus, the first input **1809** is off. In a second, switch closed state, node **1804** is at a high potential. Invertor **1807** outputs a low signal. In an embodiment, the low signal is essentially equal to ground. The potential across the first input **1809** is the difference between the supply voltage and the low signal. The potential across the first input **1809** is enough to turn on the first input. The low signal is low enough so that there is no potential across the second input **1813**. Thus, the first input **1809** is on and the second input **1813** is off in the closed switch state of circuit **1800**.

While the above embodiments described in conjunction with FIGS. **16-18** include invertors, it will be recognized that the other logic circuit elements could be used. The logic circuit elements include NAND, NOR, AND and OR gates. The use of logic elements, invertors and other logic gates, is a preferred approach as these elements use less power than the transistor switch circuit as shown in FIG. **3**.

The above embodiments described in conjunction with FIGS. **16-18** include switching between hearing aid inputs by selectively powering the inputs based on the state of a switch. It will be recognized that the switching circuits are adaptable to the other switching applications described herein. For example, the switching circuits **1600**, **1700**, or **1800** switch between an omni-directional input and a directional input.

FIG. **19** shows a hearing aid switch circuit **1900**. Circuit **1900** is similar to circuit **1600** described above with like elements being identified by reference numerals having the same two least significant digits and the two larger value digits being changed from 16 to 19. For example, the supply voltage is designated as **1601** in FIGS. **16** and **1901** in FIG. **19**. Switching circuit **1900** includes an electrical connection from the output of invertor **1907** to the signal processor **1922**. Consequently, invertor **1907** outputs a low signal to first input **1909**, second invertor **1911** and signal processor **1922** with the magnetic field sensing switch **1905** being open. Invertor **1907** outputs a high signal to first input **1909**, second invertor **1911** and signal processor **1922** with the magnetic field sensing switch **1905** being closed. Thus, the signal processor **1922** receives a hearing aid state signal from the invertor **1907**. In an embodiment, when the state signal is low, then the signal processor **1907** is adapted to optimize the hearing aid signal processing for a second (microphone) input from second input (microphone) **1913**. Second input (microphone) **1913** is in an active state as it has received a high or on signal from second invertor **1911**. The signal processing circuit **1922**, in an embodiment, optimizes the signal processing by selecting stored parameters, which are optimized for second input signal processing, from a memory. In an embodiment, the memory is an integrated circuit memory that is part of the signal processor **1922**. When the state signal is high, then the signal processor **1922** is adapted to optimize the hearing aid signal processing for a first input from first input (telecoil induction) **1909**. First input **1909** is in an active state as it has received a high or on signal from first invertor **1907**. The signal processing circuit **1922**, in an embodiment, optimizes the signal processing by selecting stored parameters, which are optimized for first input (induction) signal processing, from the memory. Other stored parameters in the memory of signal processor **1922** include automatic gain control, frequency response, and noise reduction for respective embodiments of the present disclosure.

FIG. **20** shows a hearing aid switch circuit **2000**. Circuit **2000** is similar to circuit **1700** described above with like elements being identified by reference numerals having the same two least significant digits and the two larger value digits being changed from 17 to 20. For example, the supply voltage is designated as **1701** in FIGS. **17** and **2001** in FIG. **20**. Switching circuit **2000** includes an electrical connection from the output of first invertor **2007** to the signal processor **2022**. Consequently, invertor **2007** outputs a high signal to first input **2009**, second invertor **2011** and signal processor **2022** with the magnetic field sensing switch **2005** being open. Invertor **2007** outputs a low signal to first input **2009**, second invertor **2011** and signal processor **2022** with the magnetic field sensing switch **2005** being closed. Thus, signal processor **2022** receives a hearing aid state signal from the invertor **2007**. In an embodiment, when the state signal is high, then the signal processor **2022** is adapted to optimize the hearing aid signal processing for a first input signal from first input (microphone) **2009**. First input **2009** is in an active state as it has received a high or on signal from first invertor **2007**. The signal processing circuit **2022**, in an embodiment, optimizes the signal processing by selecting stored parameters, which are optimized for microphone signal processing, from a memory. In an embodiment, the memory is an integrated circuit memory that is part of the signal processor **2022**. When the state signal is low or off, then the signal processor **2022** is adapted to optimize the hearing aid signal processing for a second input signal from second input (telecoil) **2013**. Second input **2013** is in an active state as it has received a high or on signal from second invertor **2011**. The signal processing circuit **2022**, in an embodiment, optimizes the signal processing by selecting stored parameters, which are optimized for second signal (induction) processing, from the memory. Other stored parameters in the memory of signal processor **2022** include automatic gain control, frequency response, and noise reduction for respective embodiments of the present disclosure.

FIG. **21** shows a hearing aid switch circuit **2100**. Circuit **2100** includes elements that are substantially similar to elements described above. Like elements are identified by reference numerals having the same two least significant digits and the two larger value digits being changed **21**. For example, the supply voltage is designated as **1601** in FIG. **16**, **1701** in FIGS. **17** and **2101** in FIG. **21**. Switching circuit **2100** includes a selection circuit that selects signal processing parameters. Selection circuit includes a logic gate **2107**. In the illustrated embodiment, the logic gate **2107** is a NAND gate. A first input of the NAND gate **2107** is connected to the power source **2101**. Thus, this input to the NAND gate is always high. A second input of the NAND gate **2107** is connected to the power source **2201** through a resistor and to a first terminal of magnetic field sensing switch **2105**. Consequently, the state of the switch **2105** determines the output of the NAND gate **2107** during operation of the hearing aid switch **2100**. Operation of hearing aid switch **2100** is defined as when the switch is powered. During the off or non-operational state of the hearing aid switch circuit **2100**, the supply voltage **2101** is turned off and the NAND gate **2107** will always produce a low output to conserve power, which is a consideration in designing hearing aid circuits. The switch **2105** is normally open. Thus, both inputs to the NAND gate **2107** are high and its output signal is high. The output of NAND gate **2107** is connected to signal processor **2122**. Signal processor **2122** includes a switch that upon the change of state of the NAND gate output signal changes a parameter setting in signal processor **2122**. In an embodiment, when the magnetic field sensing switch **2105** senses a magnetic field, switch **2105** closes. The second input to NAND gate **2107** goes low and NAND gate output goes low. This triggers the switch of signal processor **2122** to change parameter settings. In an embodiment, signal processor only changes its parameter settings when the signal from NAND gate **2107** shifts from high to low. In an embodiment, the parameter settings include parameters stored in a memory of signal processor **2122**. In an embodiment, a first parameter setting is adapted to process input from first input **2109**. A second parameter setting is adapted to process input from second input **2113**. In an embodiment, the first parameter setting is selected with the output signal from NAND gate **2107** being high. The second parameter setting is selected with the output signal from NAND gate **2107** being low. Accordingly, the switching circuit **2100** can select parameters that correspond to the type of input, e.g., microphone or induction inputs or directional and omni-directional inputs. The hearing aid thus more accurately produces sound for the hearing aid wearer. In an embodiment, the switch in signal processor **2122** is adapted to progress from one set of stored parameters to the next each time the signal from NAND gate **2107** goes low.

FIG. **22** shows a hearing aid switch circuit **2200**. Circuit **2200** includes elements that are substantially similar to elements described above. Like elements are identified by reference numerals having the same two least significant digits and the two larger value digits being changed **22**. For example, the supply voltage is designated as **2101** in FIG. **21** is **2201** in FIG. **22**. Switching circuit **2200** includes a selection circuit that is adapted to select parameters for signal processing. The selection circuit includes a logic gate **2207** having its output connected to signal processor **2222**. In the illustrated embodiment, the logic gate **2207** is a NAND gate. A first input of the NAND gate **2207** is connected to the power source **2201**. Thus, this input to the NAND gate is always high. A second input of the NAND gate **2207** is connected to the power source **2201** through a magnetic field sensing switch **2105**. The second input of NAND gate **2207** is also connected to ground through a resistor R. Consequently, the state of the switch **2205** determines the output of the NAND gate **2207** during operation of the hearing aid switch **2200**. Operation of hearing aid switch **2200** is defined as when the switch is powered. During the off or non-operational state of the hearing aid switch circuit **2200**, the supply voltage **2201** is turned off and the NAND gate **2207** will always produce a low output to conserve power, which is a consideration in designing hearing aid circuits. Switch **2205** is normally open. Thus, the first input to the NAND gate **2207** is high and the second input to NAND gate **2207** is low. Thus, the NAND gate output signal is low. Signal processor **2222** includes a switch that upon the change of state of the NAND gate output signal changes a parameter setting in signal processor **2222**. In an embodiment, when the magnetic field sensing switch **2205** senses a magnetic field, switch **2205** closes. The second input to NAND gate **2207** goes high and NAND gate output goes high. This triggers the switch of signal processor **2222** to change parameter settings. In an embodiment, signal processor only changes its parameter settings when the signal from NAND gate **2107** shifts from low to high. In an embodiment, the parameter settings include parameters stored in a memory of signal processor **2222**. In an embodiment, a first parameter setting is adapted to process input from first input **2209**. A second parameter setting is adapted to process input from second input **2213**. In an embodiment, the first parameter setting is selected with the output signal from NAND gate **2207** being low. The second parameter setting is selected with the output signal from NAND gate **2207** being high. Accordingly, the switching circuit **2200** can select parameters that correspond to the type of input, e.g., microphone or induction inputs. The hearing aid thus more accurately produces sound for the hearing aid wearer.

It will be appreciated that the selection of parameters for specific inputs can be combined with the FIGS. **2-18** embodiments. For example, the magnetic field sensor changing state not only switches the input but also generates a signal, for example, through logic circuit elements, that triggers the signal processing circuit to change its operational parameters to match the type of input.

Possible applications of the technology include, but are not limited to, hearing aids. Various types of magnetic field sensors are described herein for use in hearing aids. One type is a mechanical reed switch. Another type is a solid state magnetic responsive sensor. Another type is a MEMS switch. Another type is a GMR sensor. Another type is a core saturation circuit. Another type is anisotropic magneto resistive circuit. Another type is magnetic field effect transistor. It is desirable to incorporate solid state devices into hearing aids as solid state devices typically are smaller, consume less power, produce less heat then discrete components. Further the solid state switching devices can sense and react to a varying magnetic field at a sufficient speed so that the magnetic field is used for supplying programming signals to the hearing aid.

Those skilled in the art will readily recognize how to realize different embodiments using the novel features of the present invention. Several other embodiments, applications and realizations are possible without departing from the present invention. Consequently, the embodiment described herein is not intended in an exclusive or limiting sense, and that scope of the invention is as claimed in the following claims and their equivalents.

What is claimed is:

1. A hearing aid, comprising:

an input system;

an output system;

a solid state tunneling magnetic sensor generating a magnetic field signal;

a processor configured to be programmed to process signals from the input system and provide the processed signals to the output system,

wherein the processor is configured to receive the magnetic field signal from the sensor, and is programmable to select parameters for signal processing using a first digital filter or a second digital filter, the selection of either the first digital filter or the second digital filter based at least in part on the magnetic field signal.

2. The hearing aid of claim **1**, wherein the solid state tunneling magnetic sensor includes a spin dependent tunneling (SDT) device.

3. The hearing aid of claim **2**, wherein the SDT device is fabricated using photolithography.

4. The hearing aid of claim **2**, wherein the SDT device includes a saturation field range from 0.1 to 10 kA/m.

5. The hearing aid of claim **2**, wherein the SDT device is configured to be used as a hearing aid switch.

6. The hearing aid of claim **2**, wherein the SDT device is configured to provide hearing aid programming signals.

7. The hearing aid of claim **2**, wherein the SDT device includes a giant magnetoresistivity (GMR) material layer, and wherein the SDT device includes a conduction path perpendicular to a plane of the GMR material layer.

8. The hearing aid of claim **1**, wherein the input system includes a microphone.

9. The hearing aid of claim **1**, wherein the input system is configured to switch from an acoustic input to a magnetic input based on the magnetic field signal.

10. The hearing aid of claim **9**, wherein the magnetic input includes a telecoil.

11. A hearing aid, comprising:

a power source;

a hearing aid circuit;

a solid state tunneling magnetic sensor configured to connect the power source to the hearing aid circuit, wherein the sensor is configured to disconnect the power source from the hearing aid circuit when in the presence of a sufficiently strong magnetic field; and

wherein the solid state tunneling magnetic sensor includes a spin dependent tunneling (SDT) device.

12. The hearing aid of claim **11**, wherein the SDT device is fabricated using photolithography.

13. The hearing aid of claim **11**, wherein the SDT device includes a saturation field range from 0.1 to 10 kA/m.

14. The hearing aid of claim **11**, wherein the SDT device includes a giant magnetoresistivity (GMR) material layer, and wherein the SDT device includes a conduction path perpendicular to a plane of the GMR material layer.

15. The hearing aid of claim **11**, wherein the power source is a battery.

16. The hearing aid of claim **15**, wherein the battery is rechargeable.

17. The hearing aid of claim **11**, further comprising a filter connected to the hearing aid circuit, wherein the solid state tunneling magnetic sensor is configured to electrically disconnect the filter from the hearing aid circuit when in the presence of a sufficiently strong magnetic field.

18. The hearing aid of claim **11**, wherein the solid state tunneling magnetic sensor is further configured to operate as a programming circuit to program the hearing aid.

19. The hearing aid of claim **11**, further comprising at least one acoustic input connected to the hearing aid circuit, wherein the solid state tunneling magnetic sensor is configured to inhibit the acoustic input in the presence of a magnetic field.

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UNITED STATES PATENT AND TRADEMARK OFFICE

CERTIFICATE OF CORRECTION

PATENT NO. : 8,971,559 B2
APPLICATION NO. : 13/873031
DATED : March 3, 2015
INVENTOR(S) : Sacha et al.

It is certified that error appears in the above-identified patent and that said Letters Patent is hereby corrected as shown below:

Title Page, item (56)

On page 4, in column 2, under “Other Publications”, line 47, delete “Feb. 20,” and insert --Mar. 20,--, therefor

On page 5, in column 2, under “Other Publications”, line 17, delete “Conncetivity” and insert --Connectivity--, therefor

Signed and Sealed this
Second Day of February, 2016

Michelle K. Lee

Michelle K. Lee
Director of the United States Patent and Trademark Office